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(54) **MICROFLUIDIC ROTARY FLOW REACTOR MATRIX**

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(58) **Field of Classification Search** **422/100, 422/99, 50, 63, 500; 137/824; 435/6, 7.9, 435/287.2**

See application file for complete search history.

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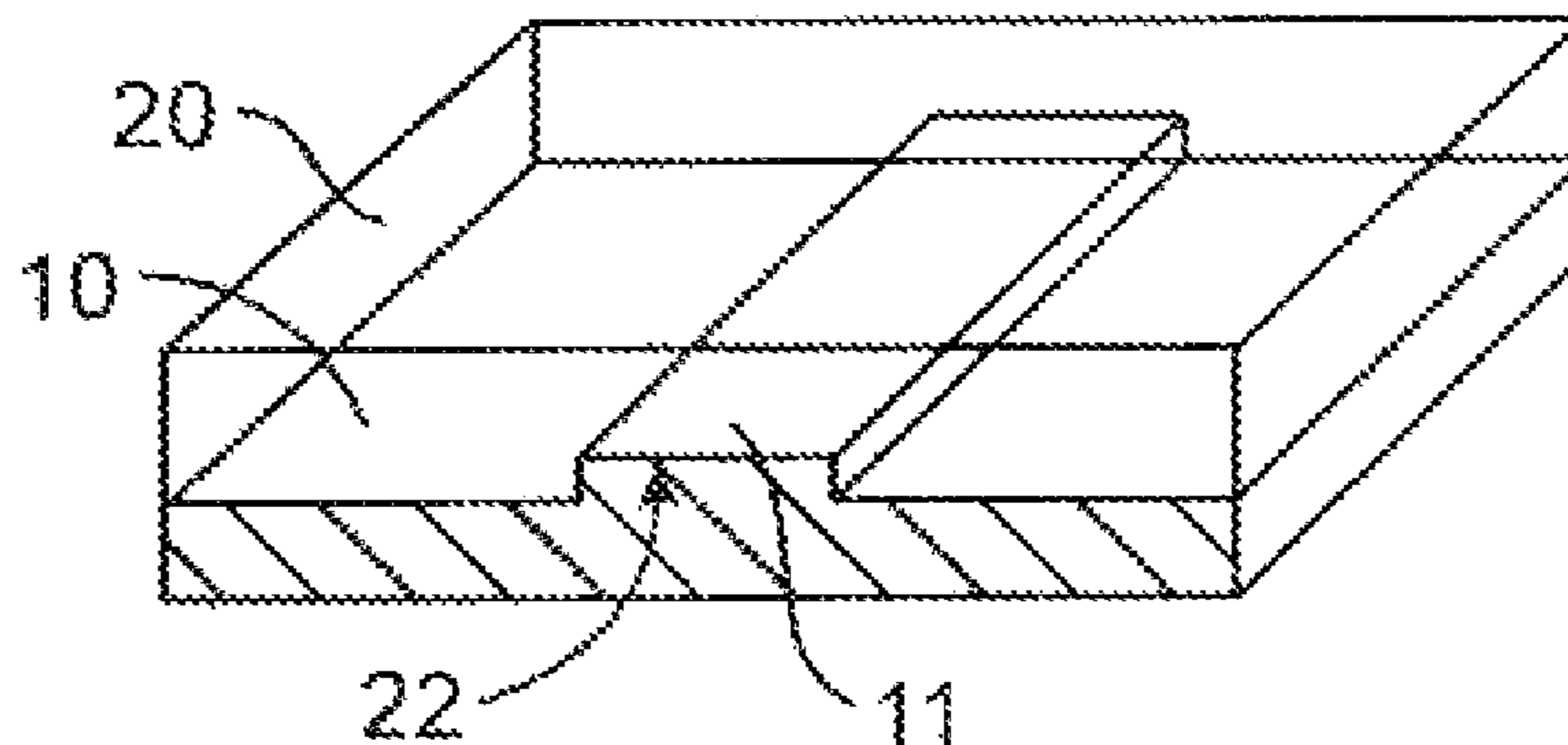
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(57) **ABSTRACT**

A microfluidic device comprises a matrix of rotary flow reactors. The microfluidic matrix device offers a solution to the “world-to-chip” interface problem by accomplishing two important goals simultaneously: an economy of scale in reagent consumption is achieved, while simultaneously minimizing pipetting steps. N^2 independent assays can be performed with only $2N+1$ pipetting steps, using a single aliquot of enzyme amortized over all reactors. The chip reduces labor relative to conventional fluid handling techniques by using an order of magnitude less pipetting steps, and reduces cost by consuming two to three orders of magnitude less reagents per reaction. A PCR format has immediate applications in medical diagnosis and gene testing. Beyond PCR, the microfluidic matrix chip provides a universal and flexible platform for biological and chemical assays requiring parsimonious use of precious reagents and highly automated processing.

9 Claims, 38 Drawing Sheets



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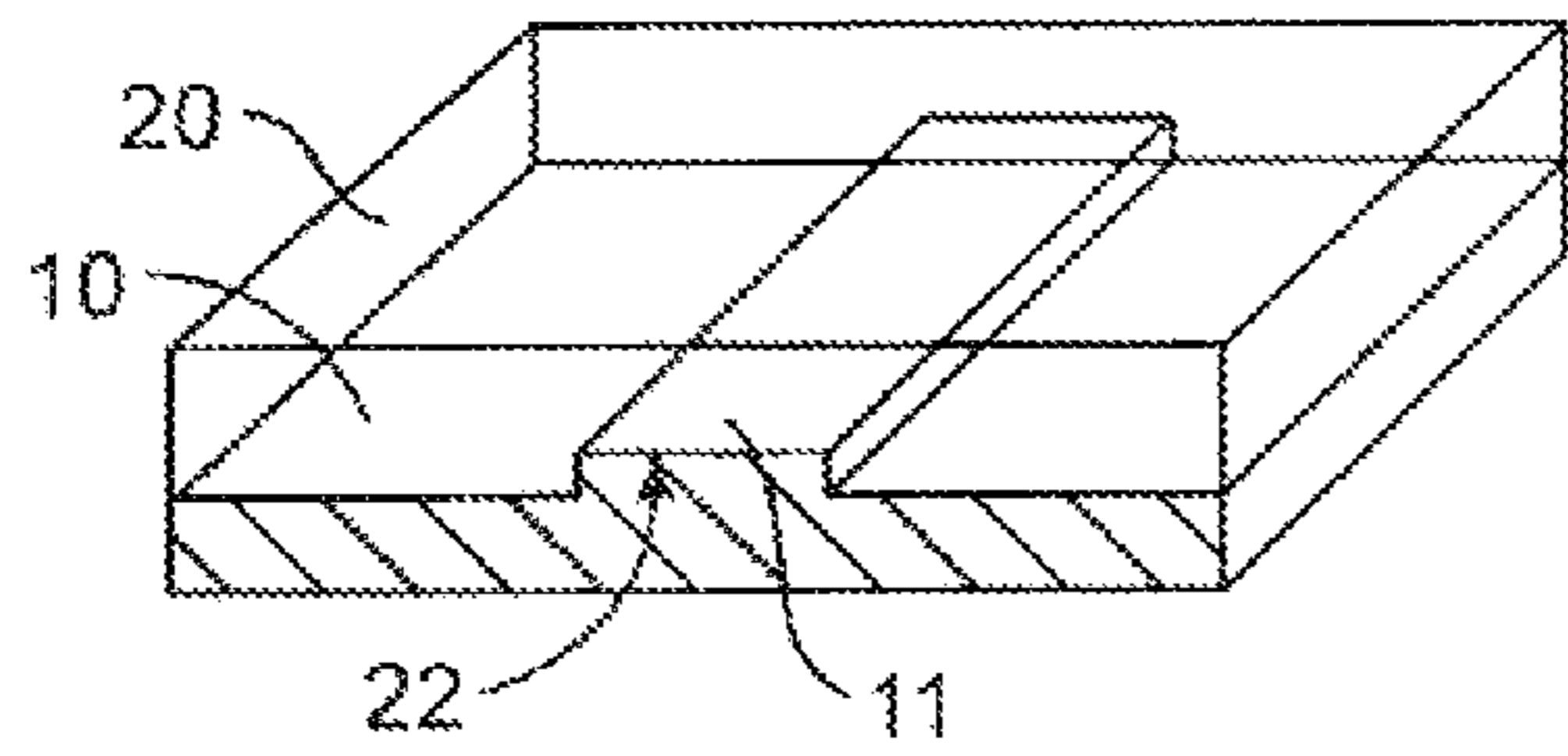


FIG. 1

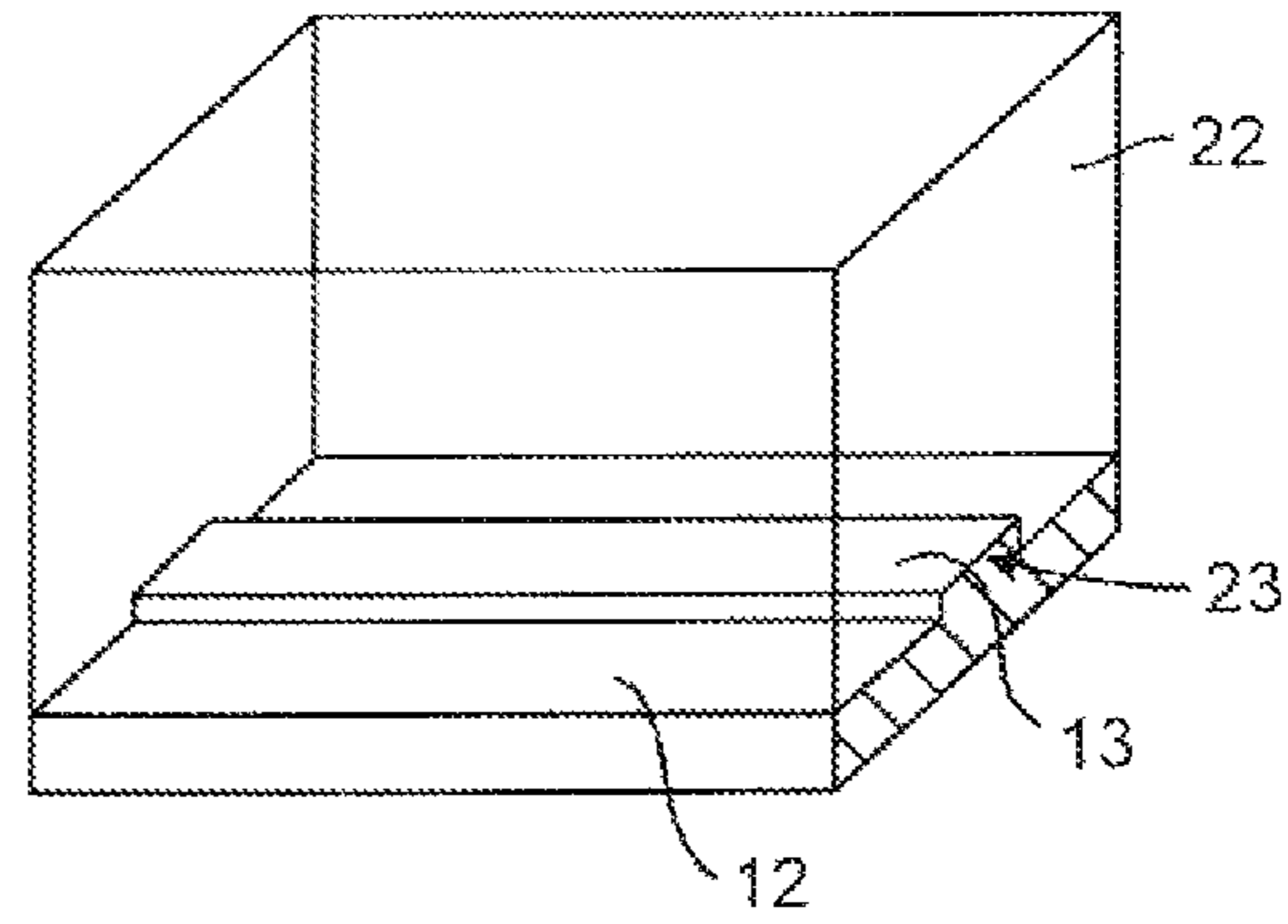


FIG. 2

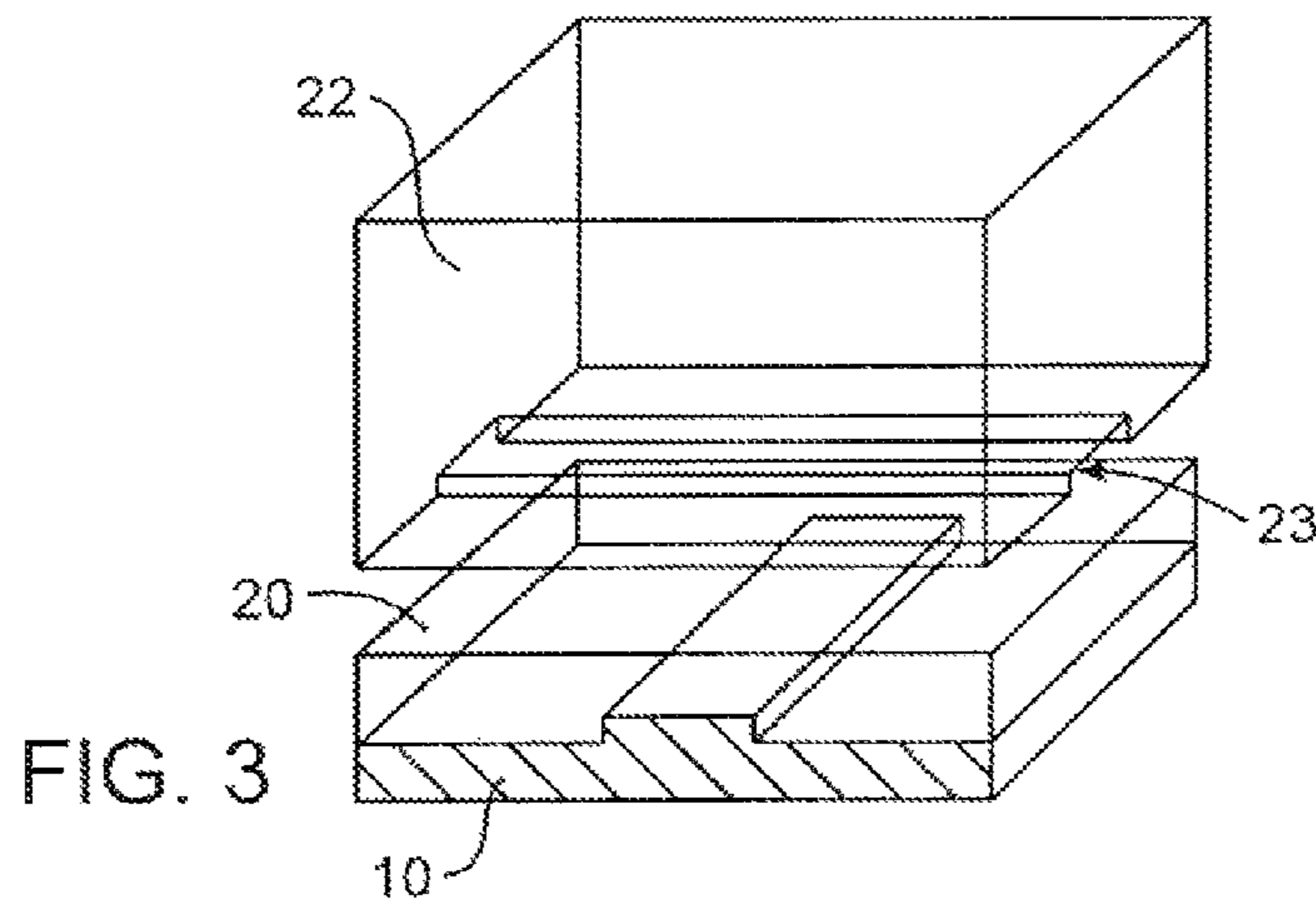


FIG. 3

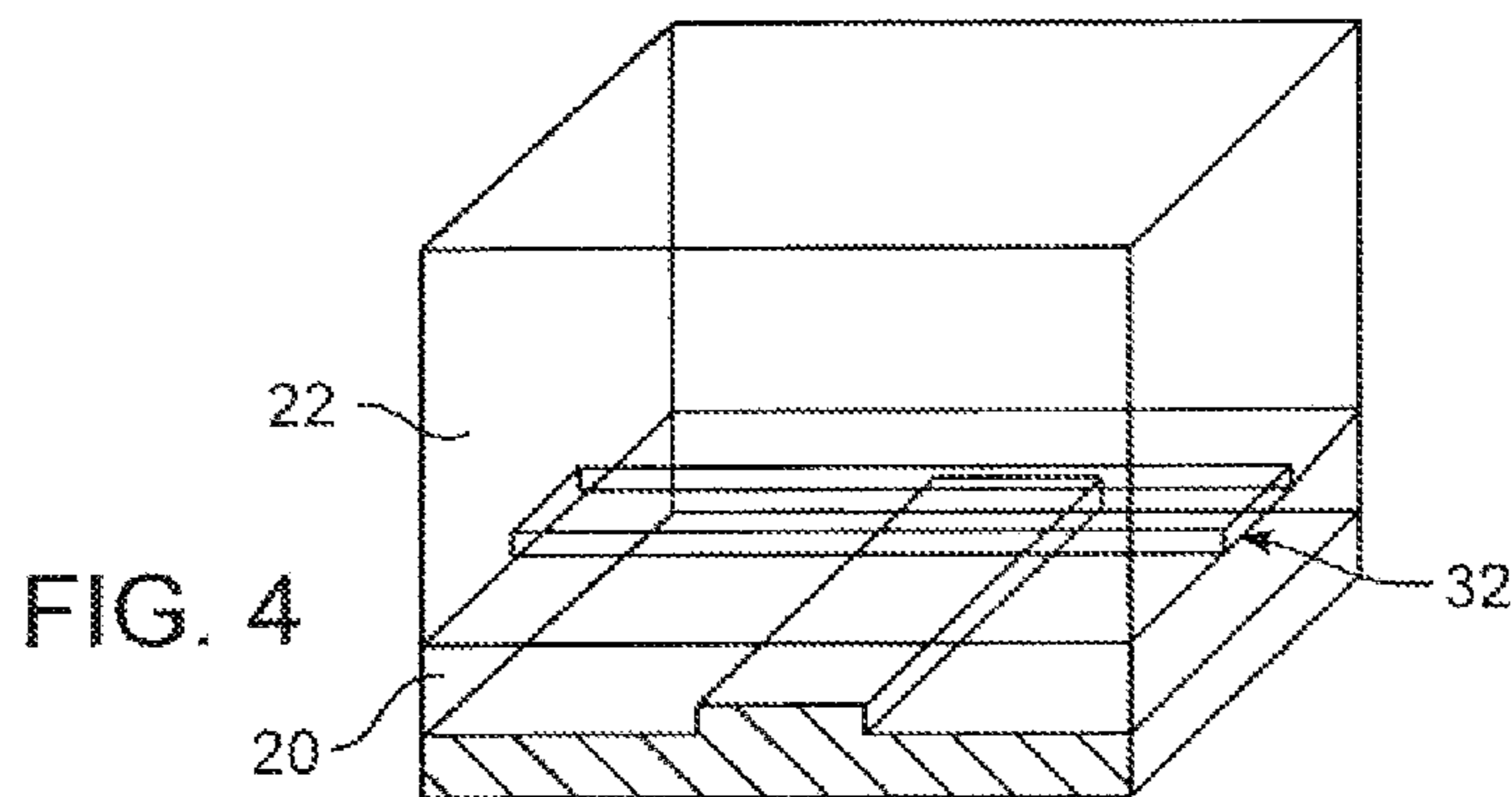
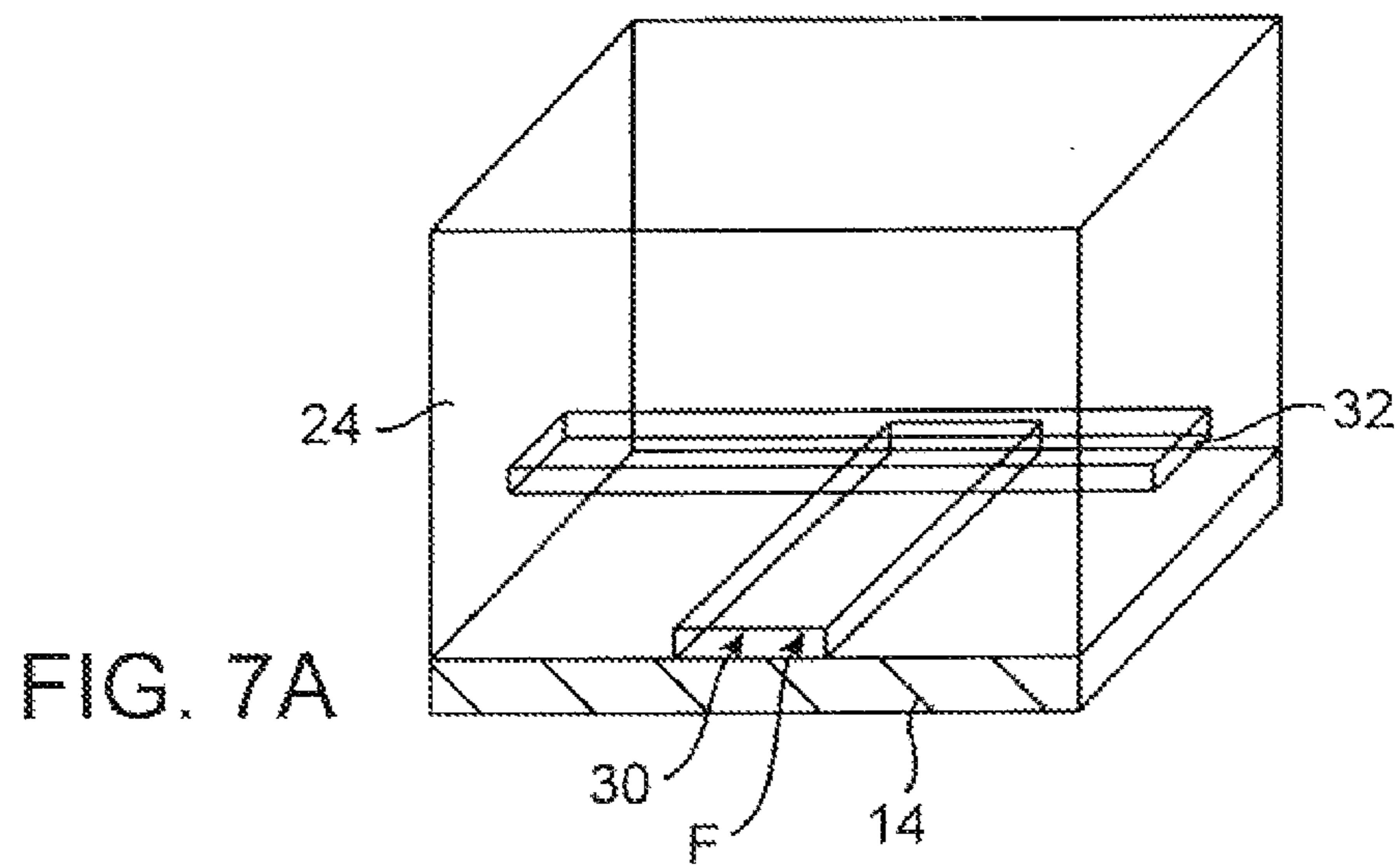
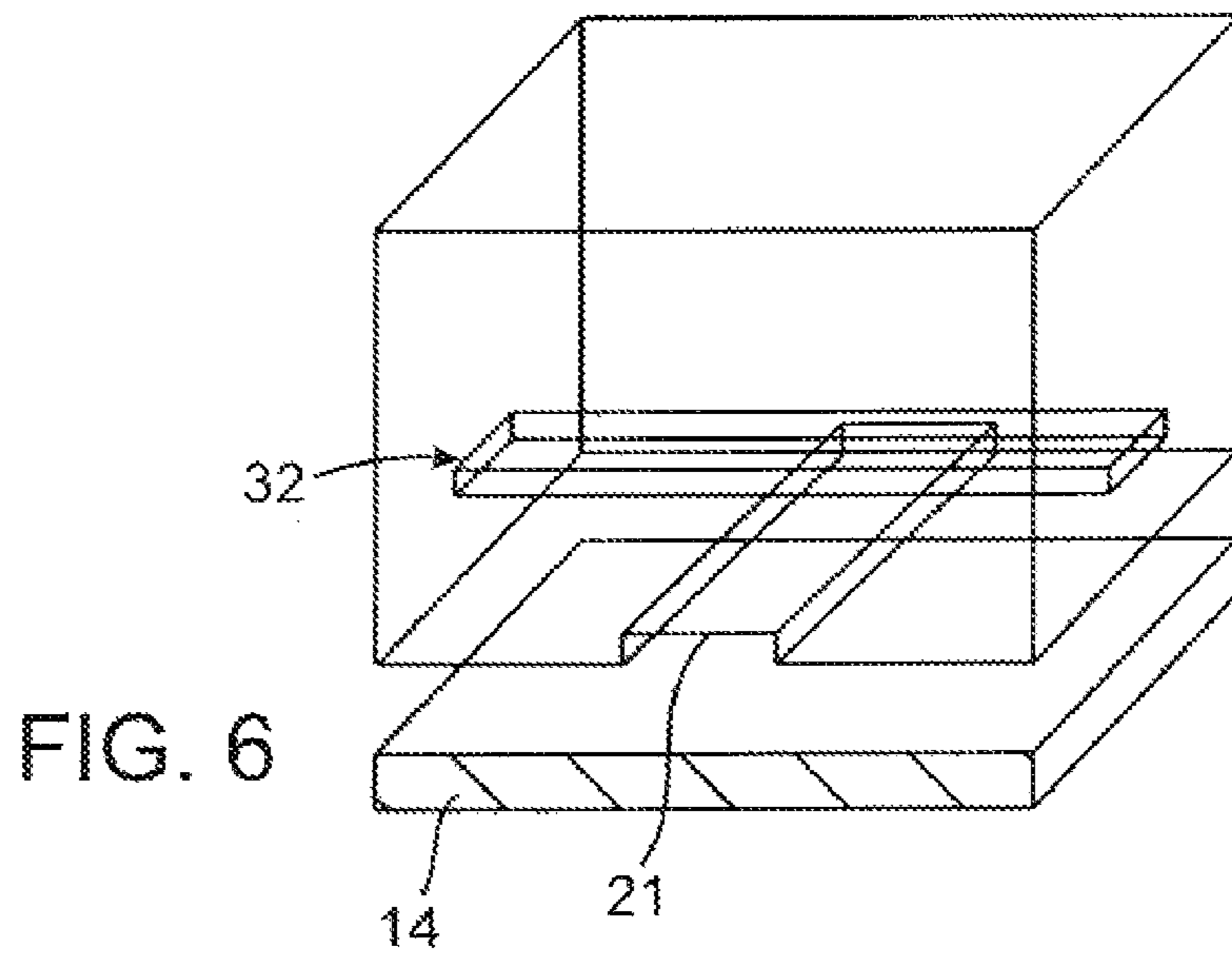
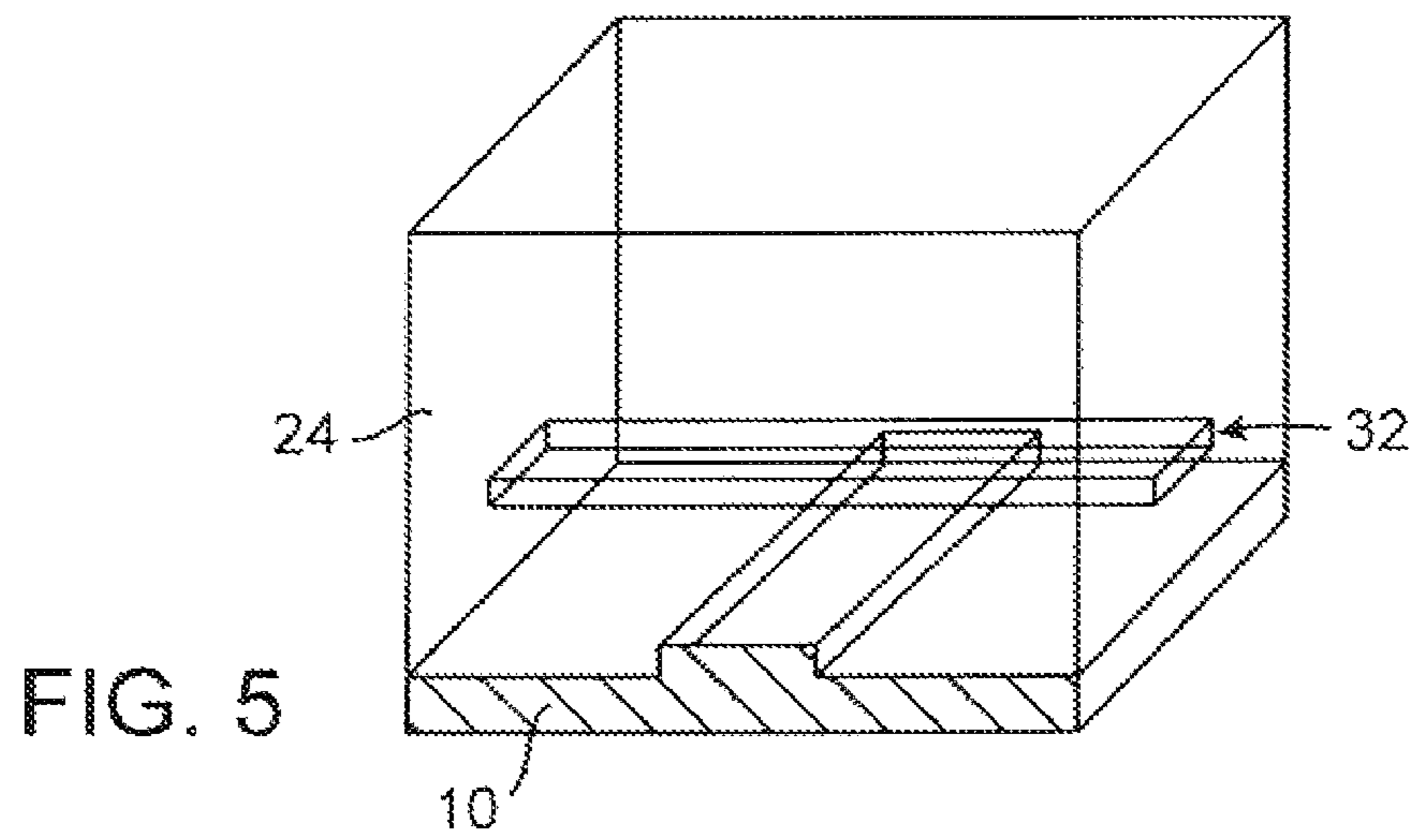


FIG. 4



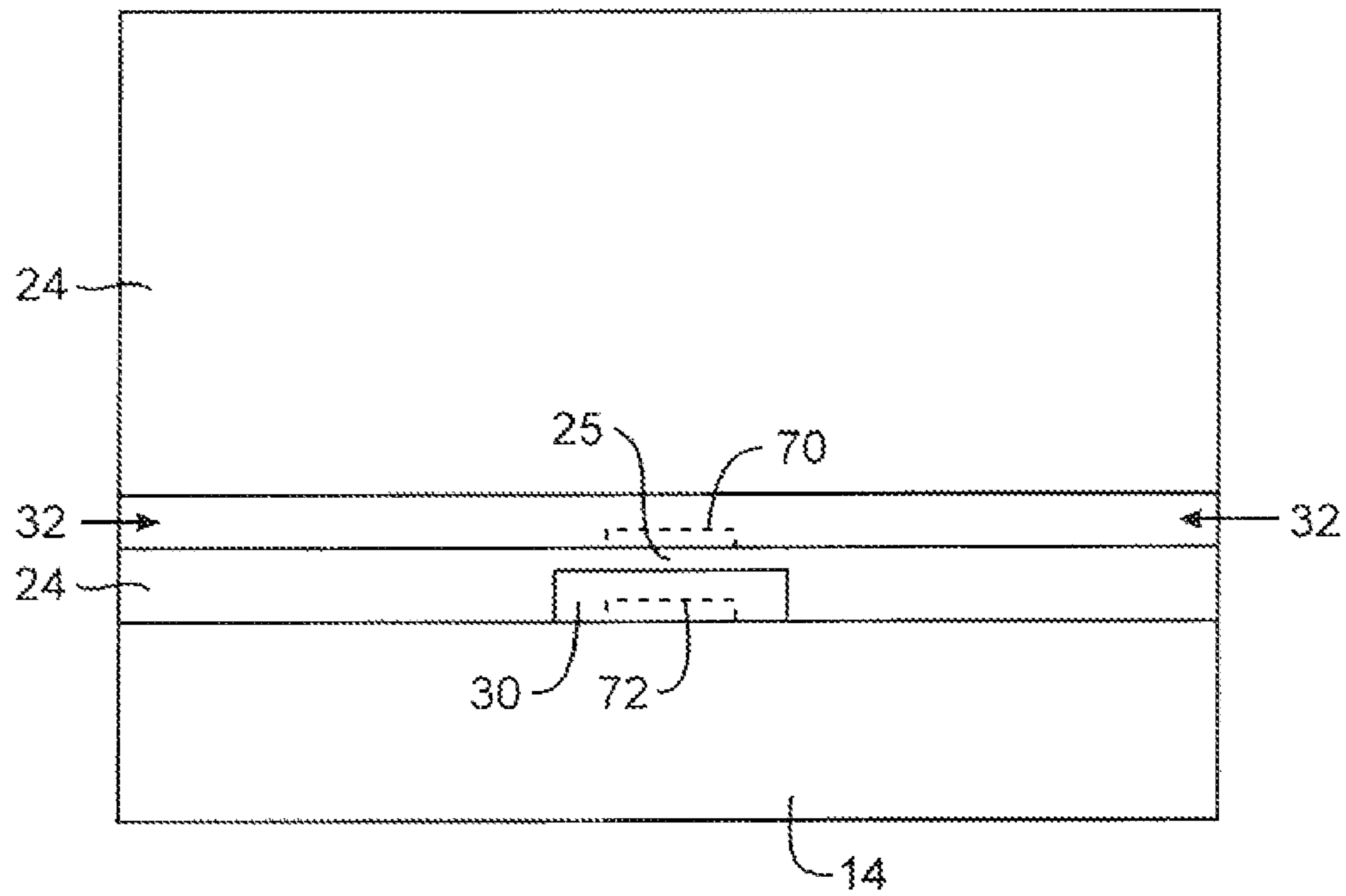


FIG. 7B

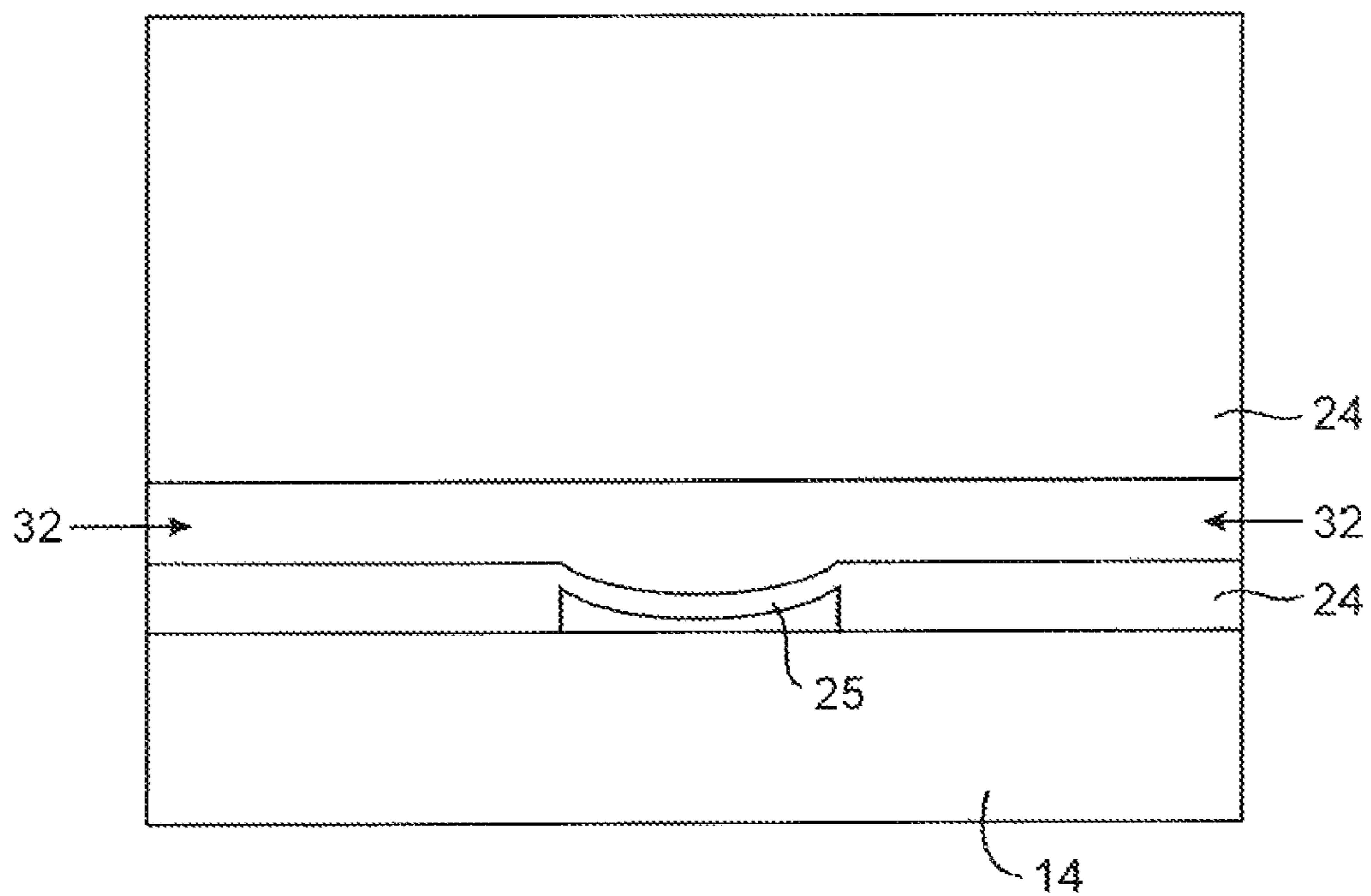
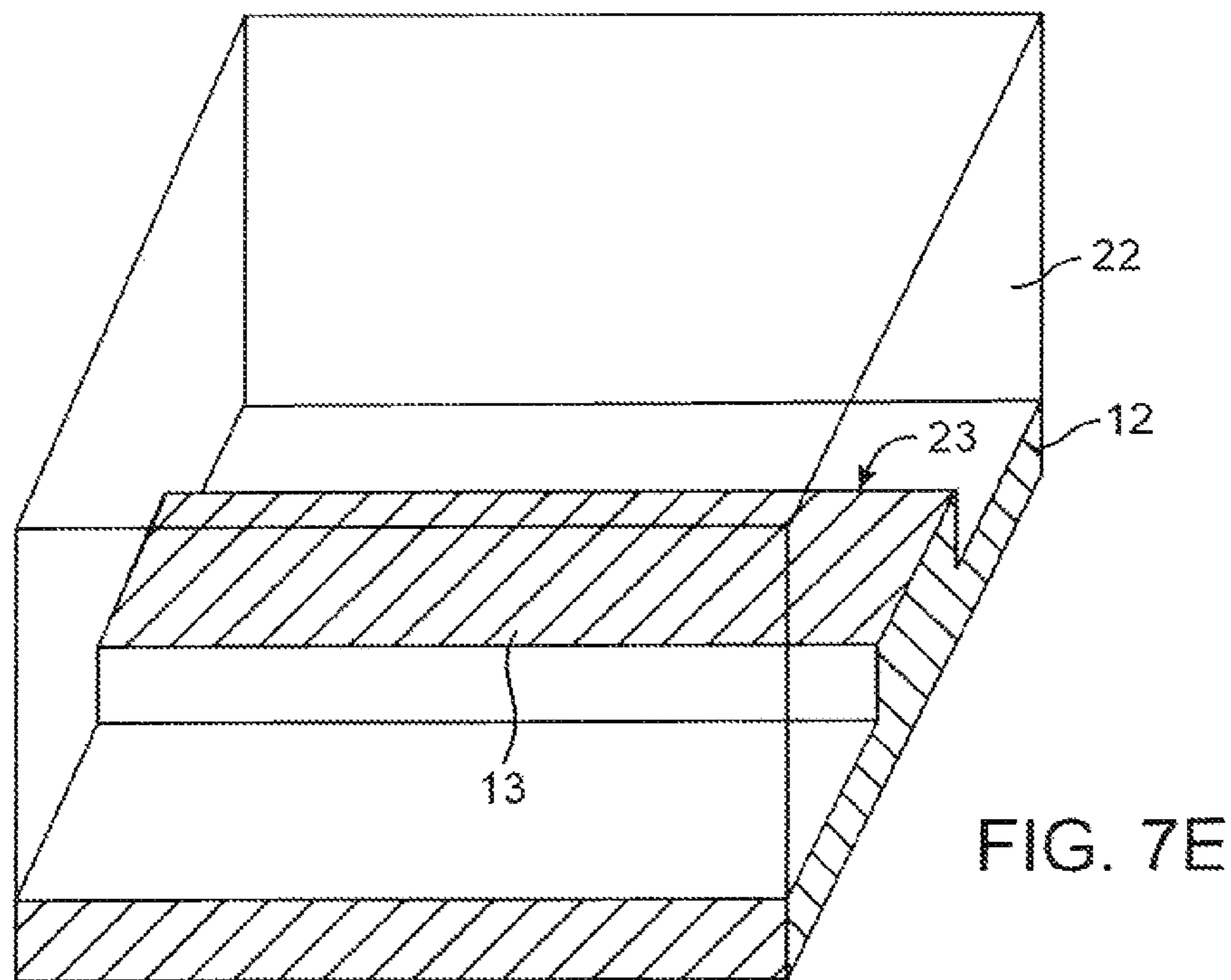
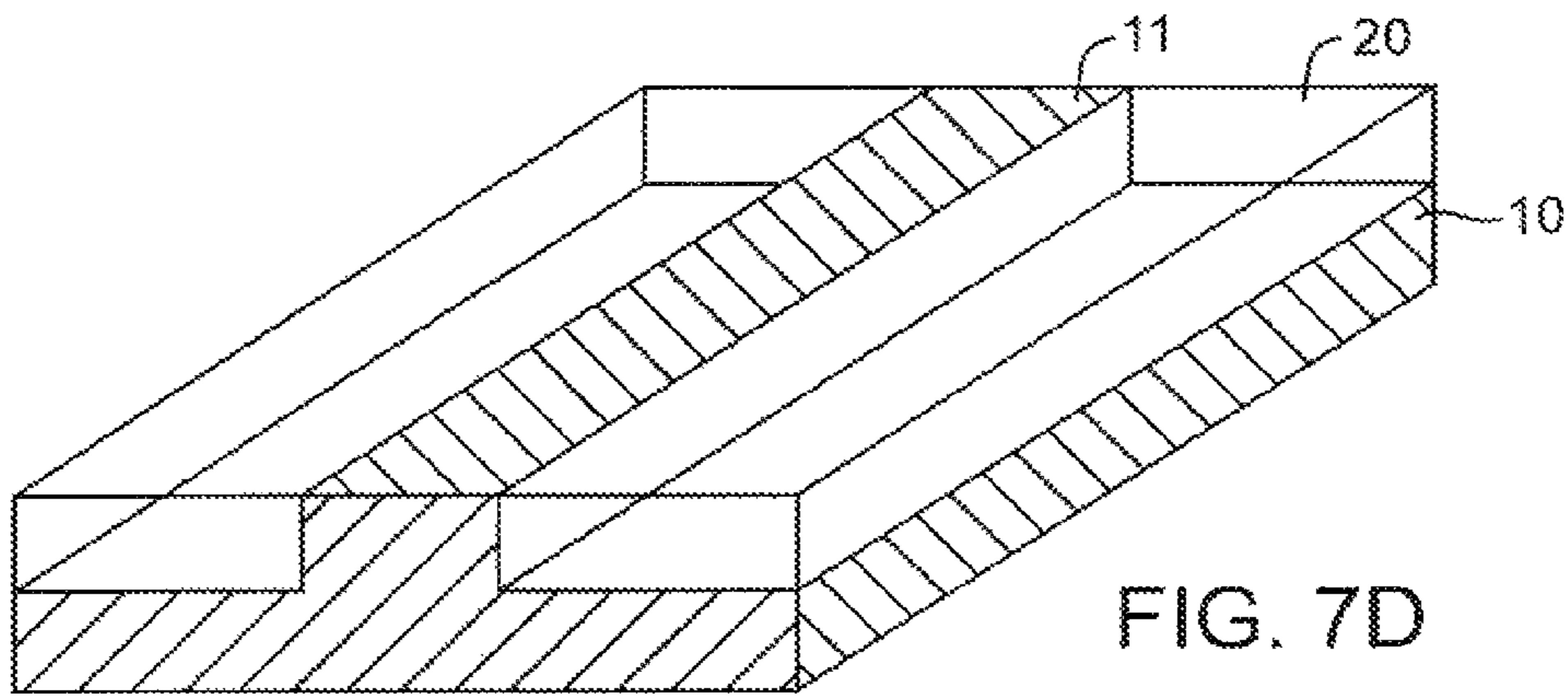
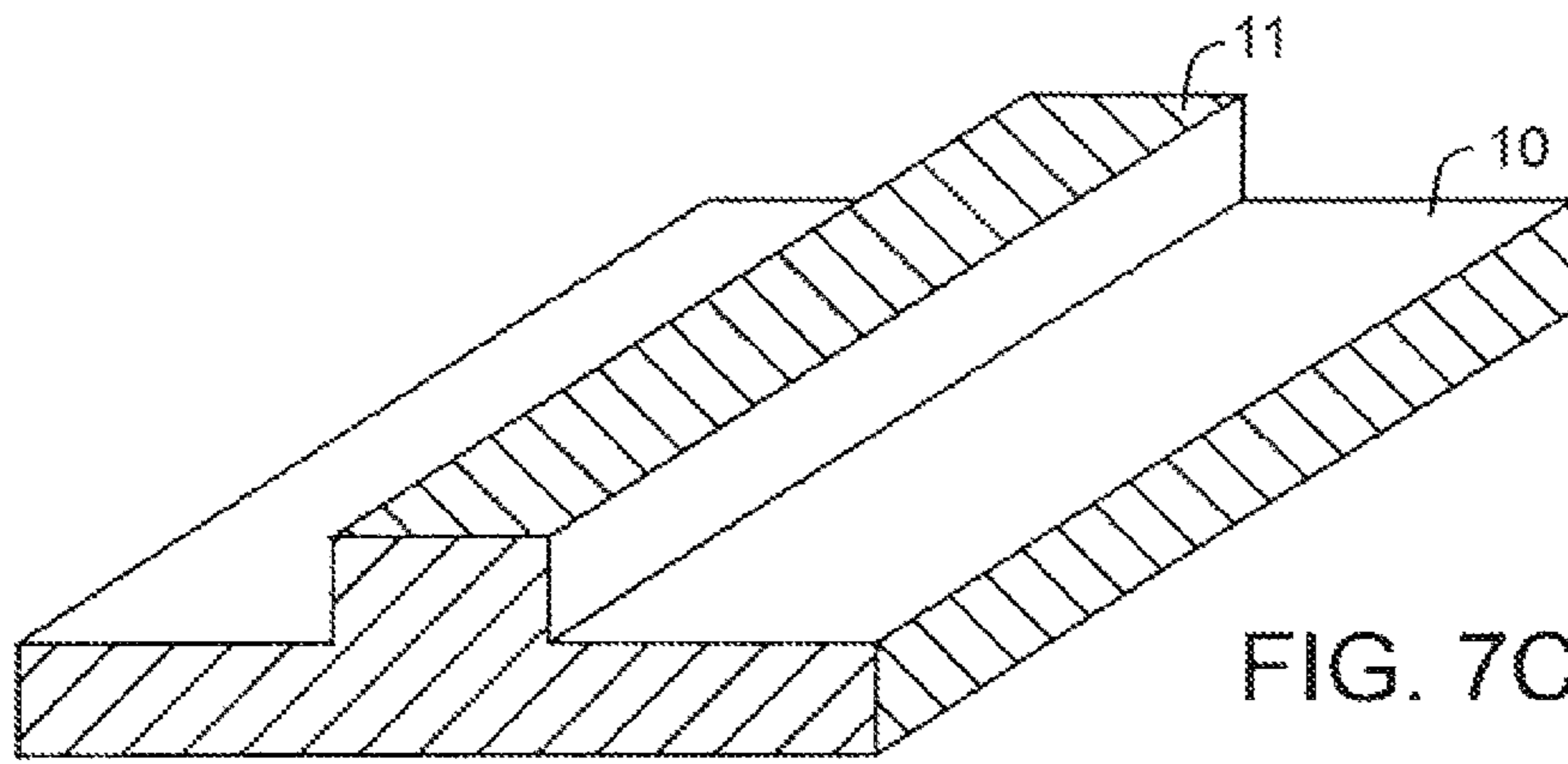
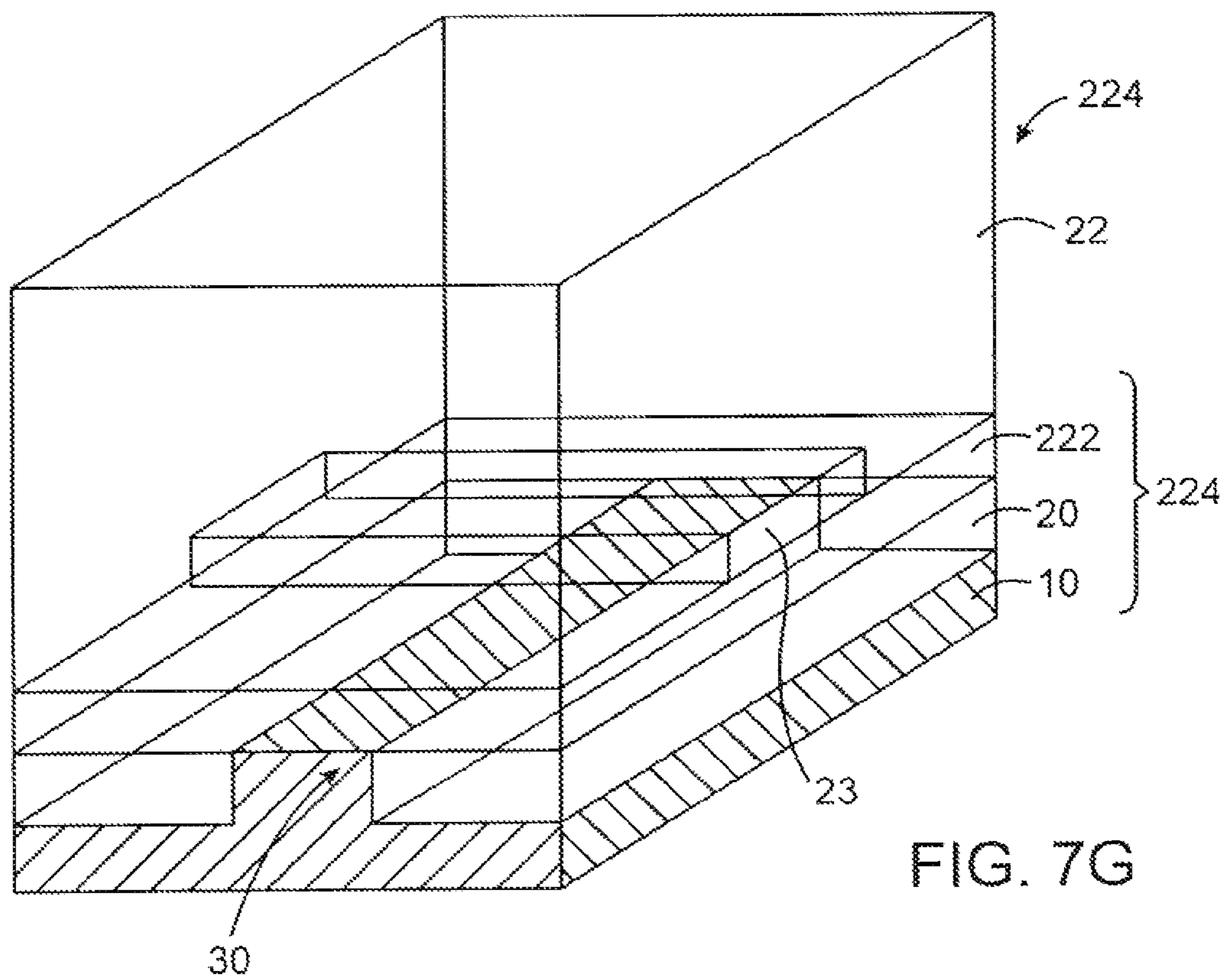
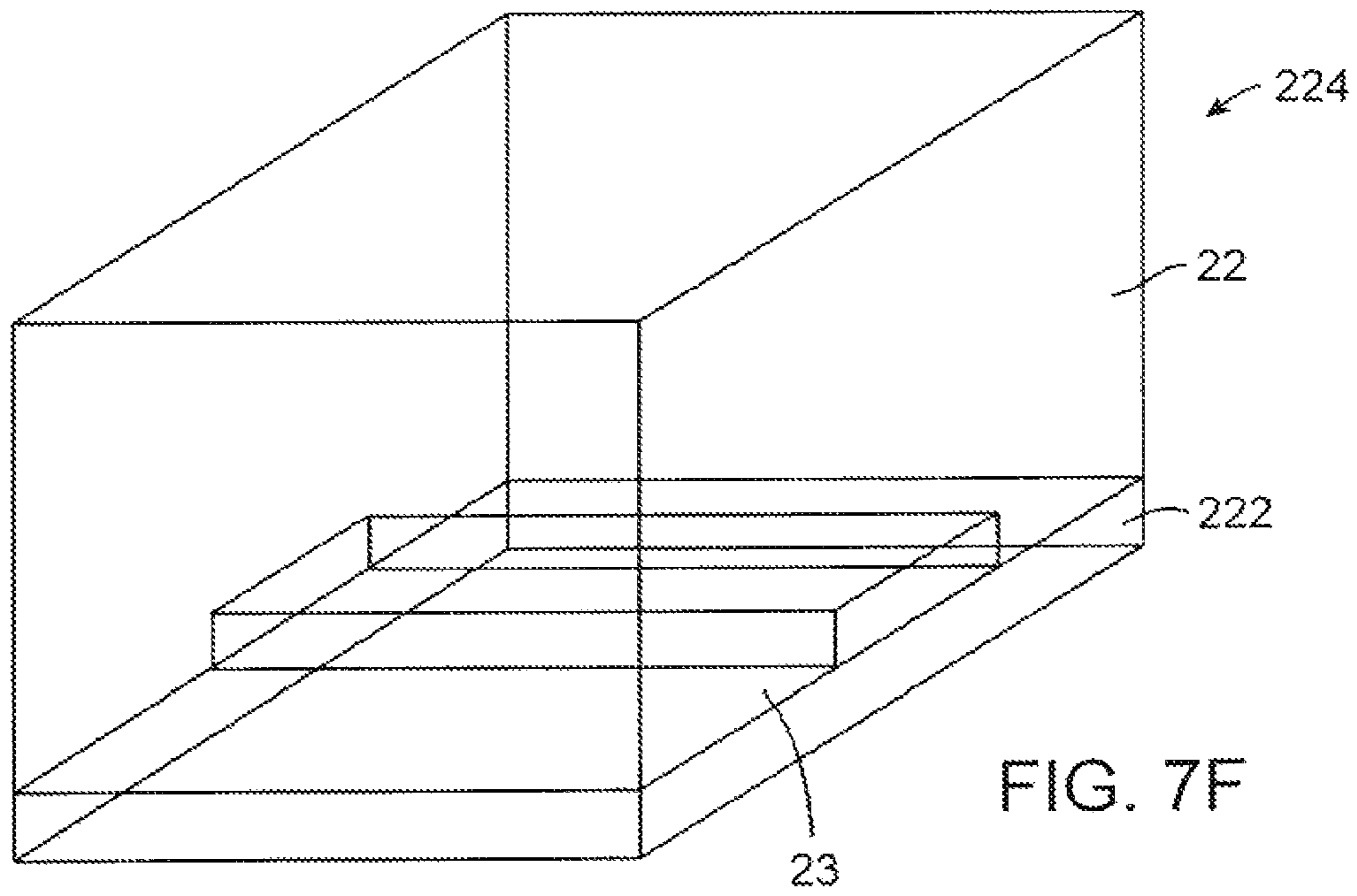


FIG. 7H





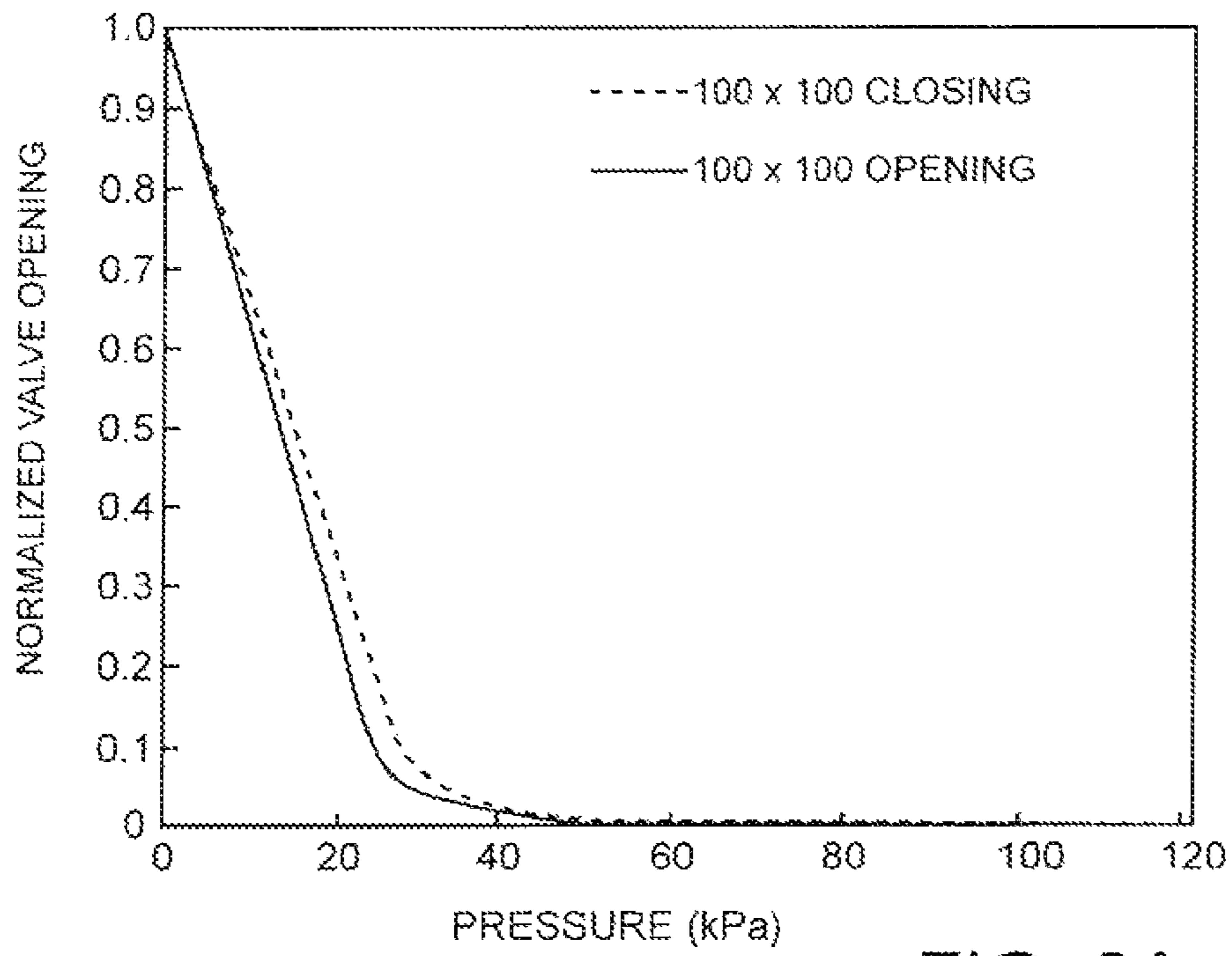


FIG. 8A

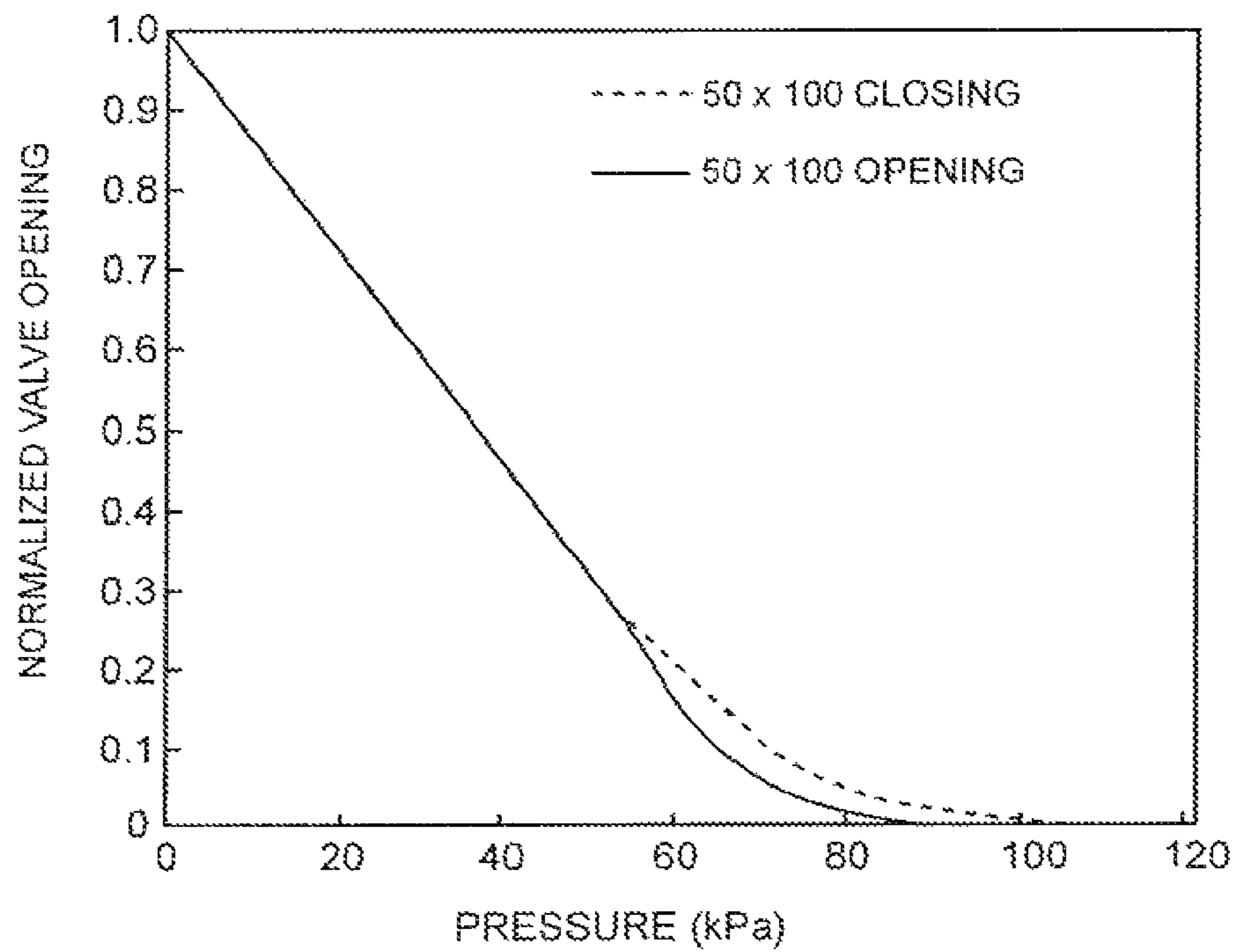


FIG. 8B

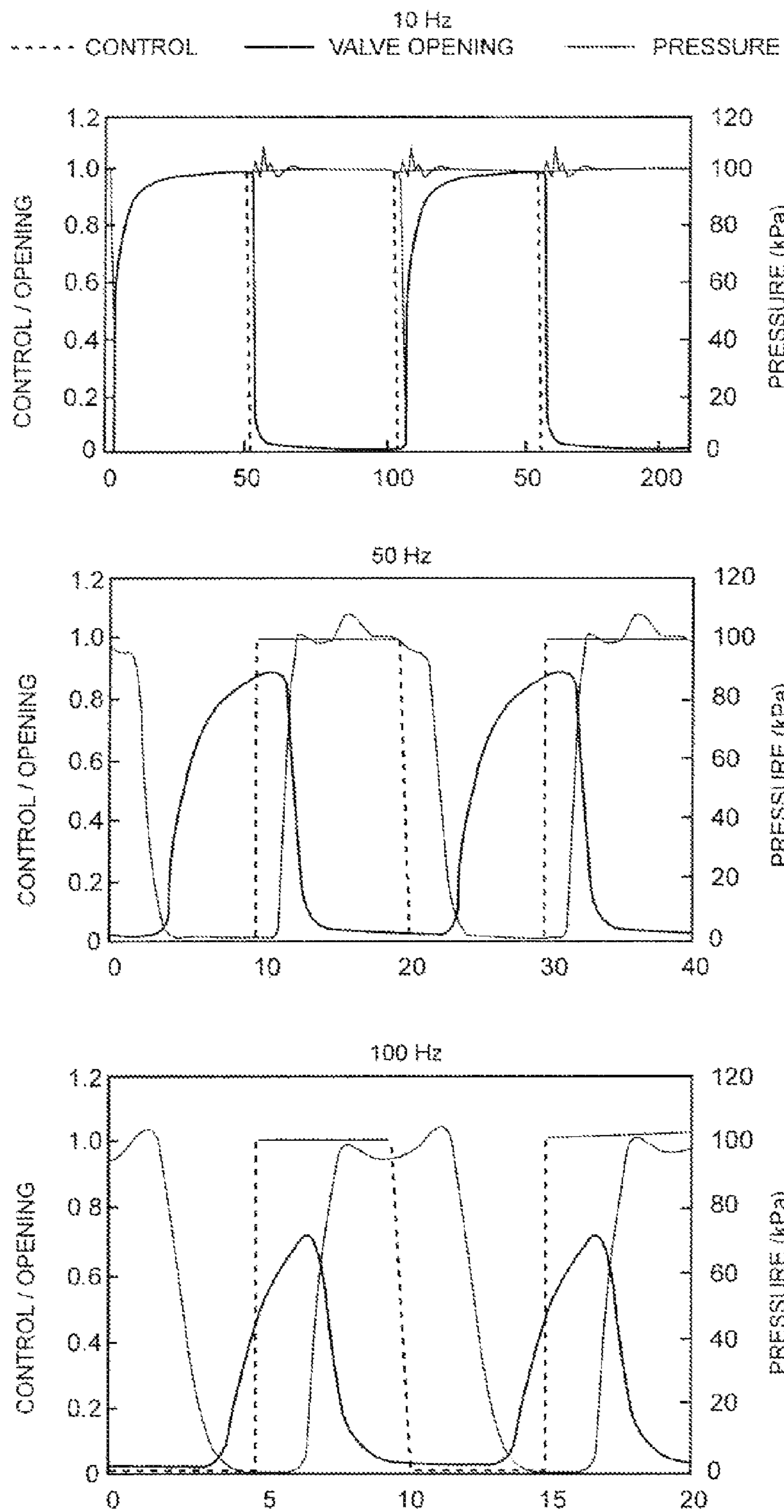


FIG. 9

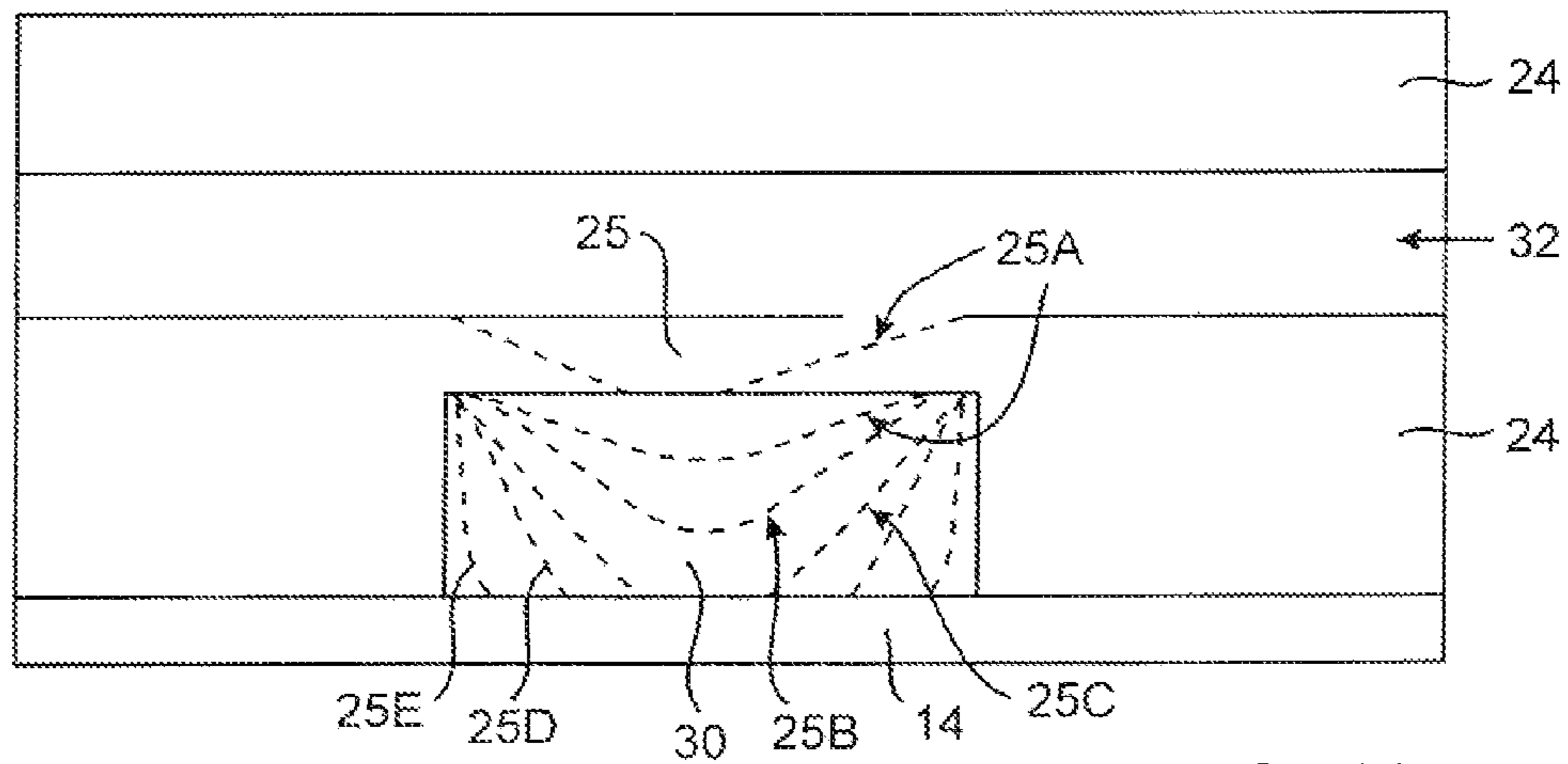


FIG. 10

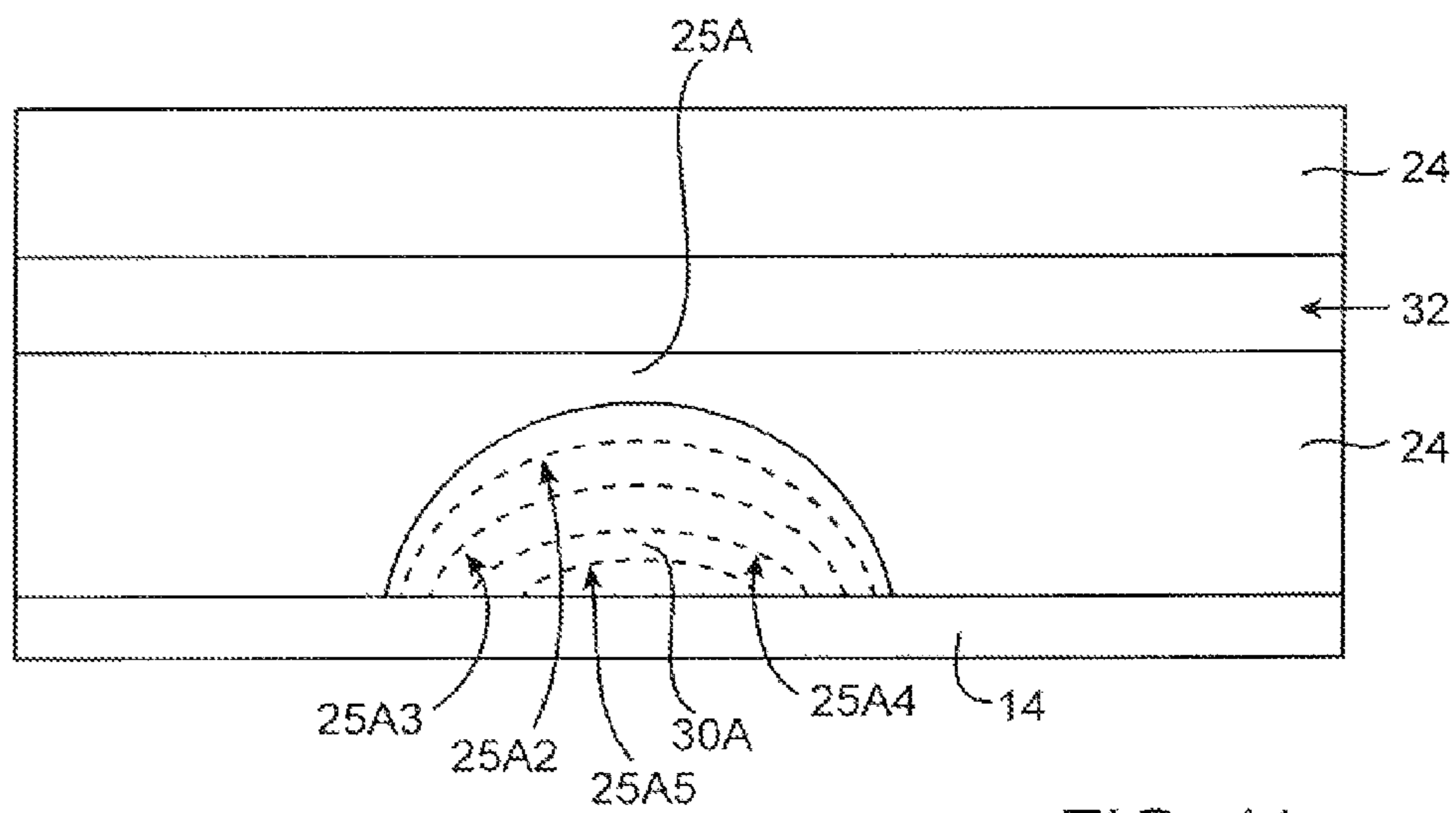


FIG. 11

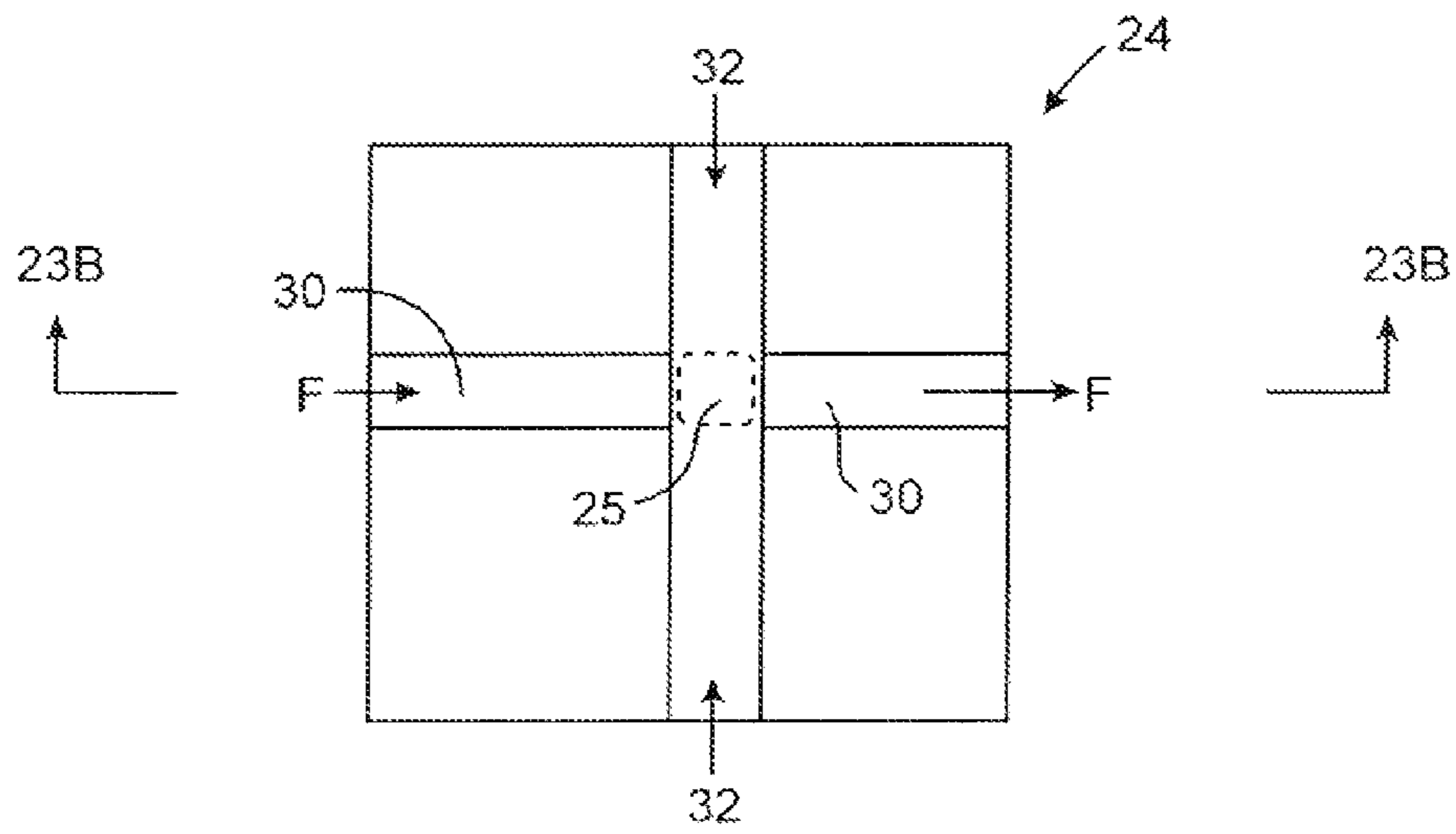


FIG. 12A

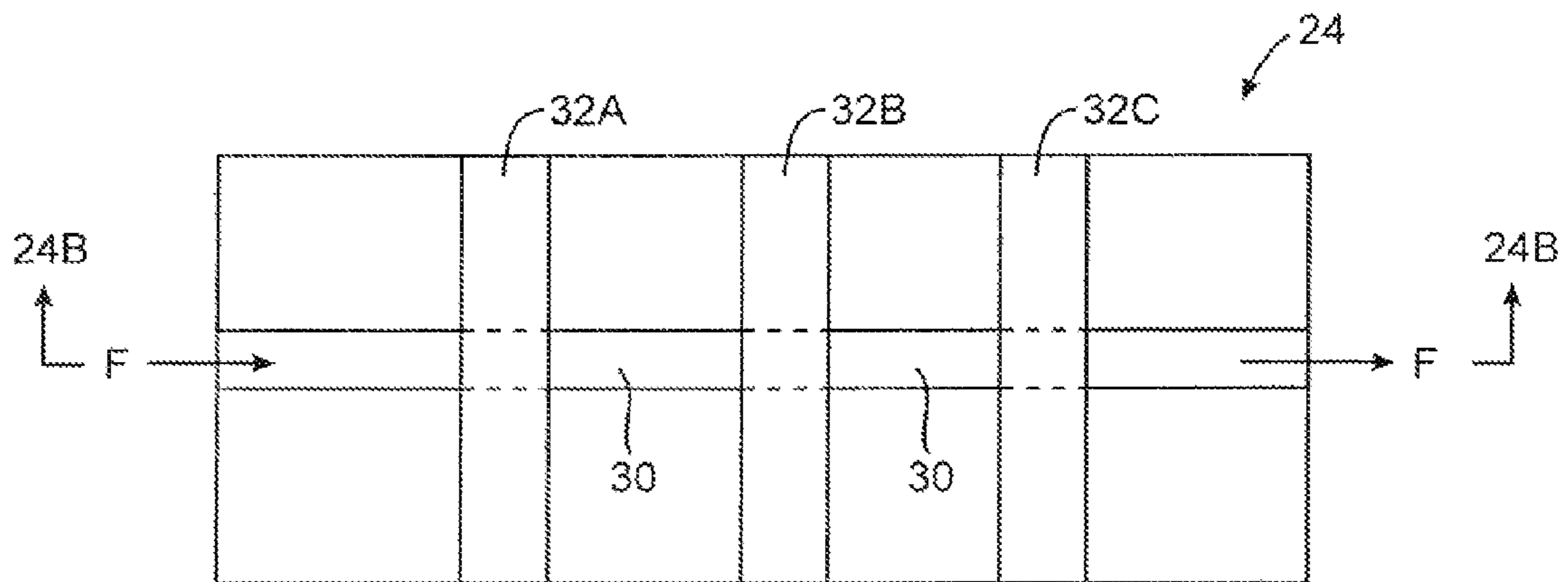


FIG. 13A

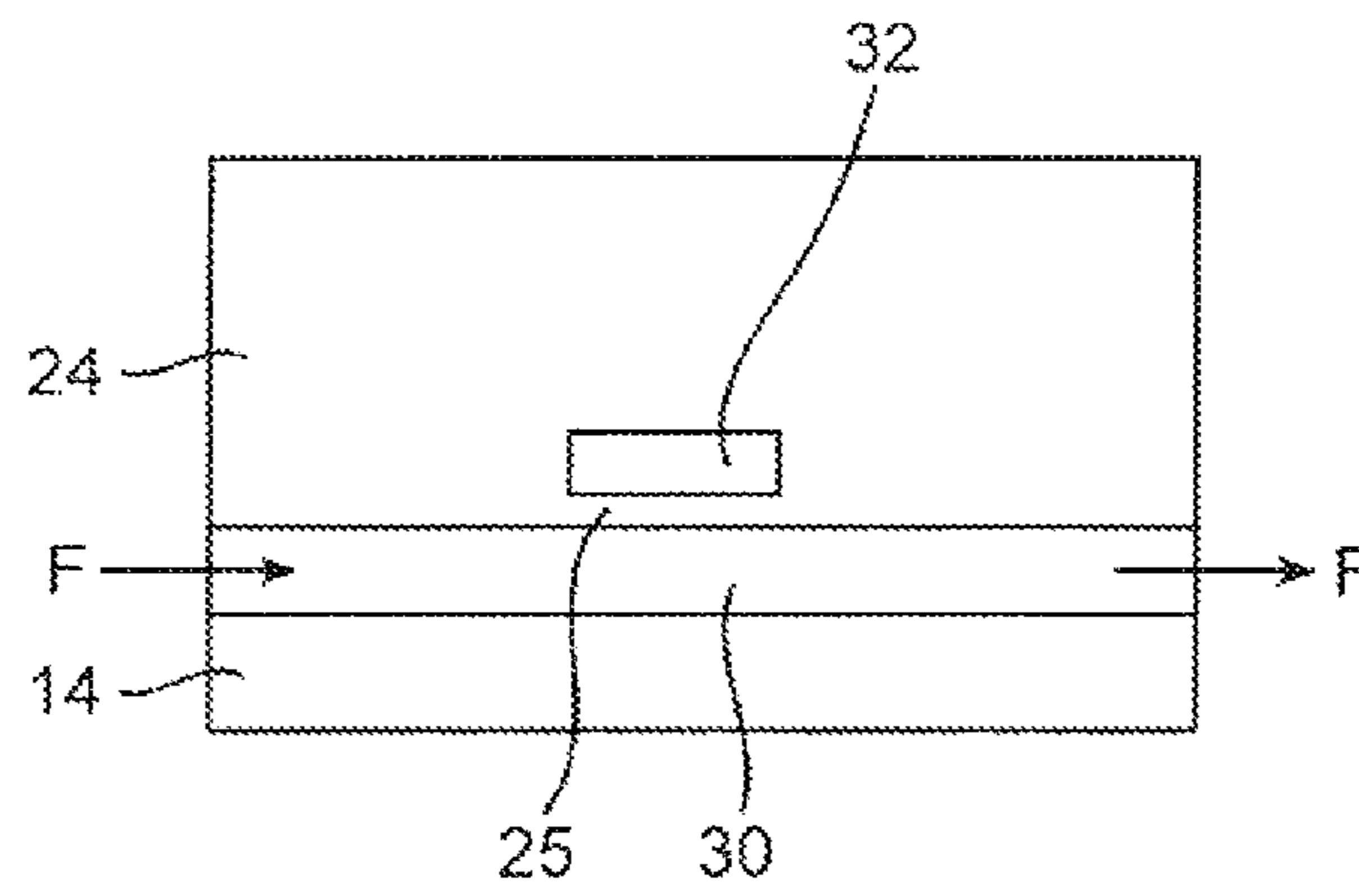


FIG. 12B

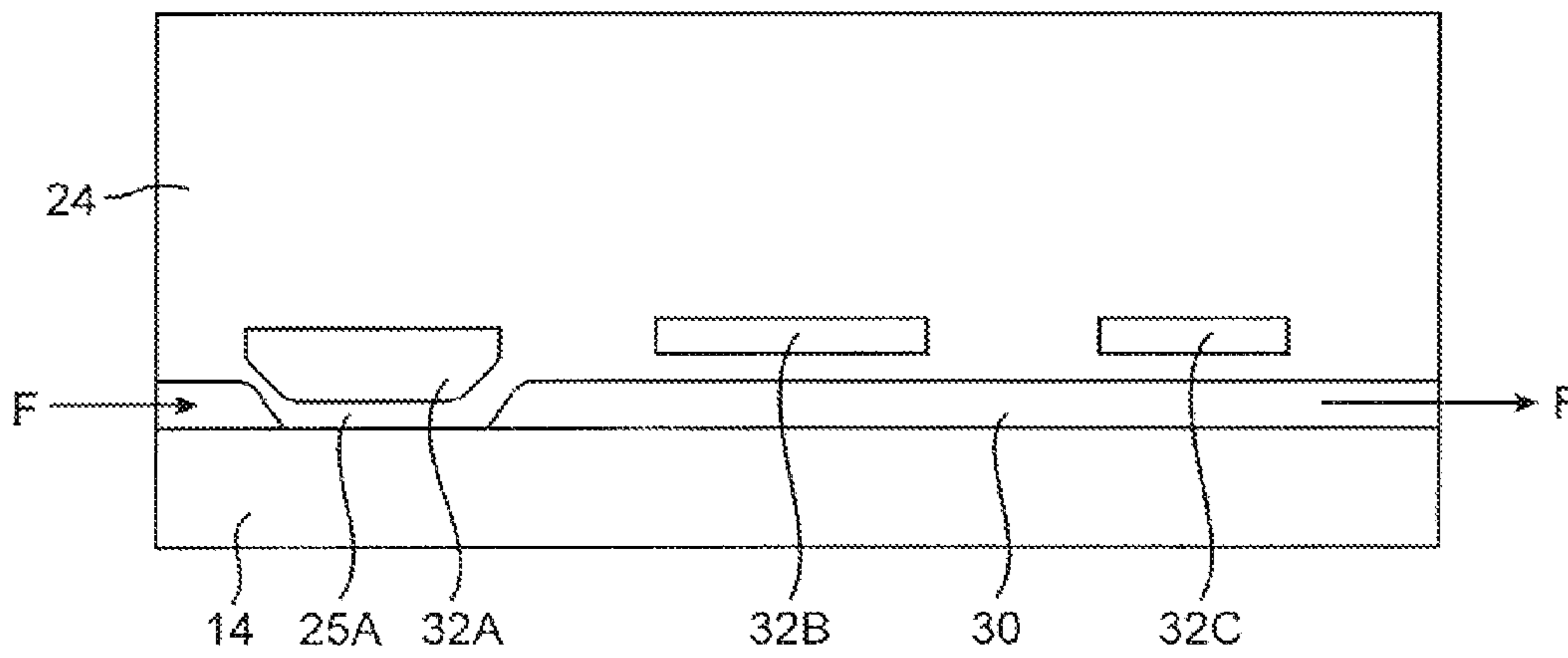


FIG. 13B

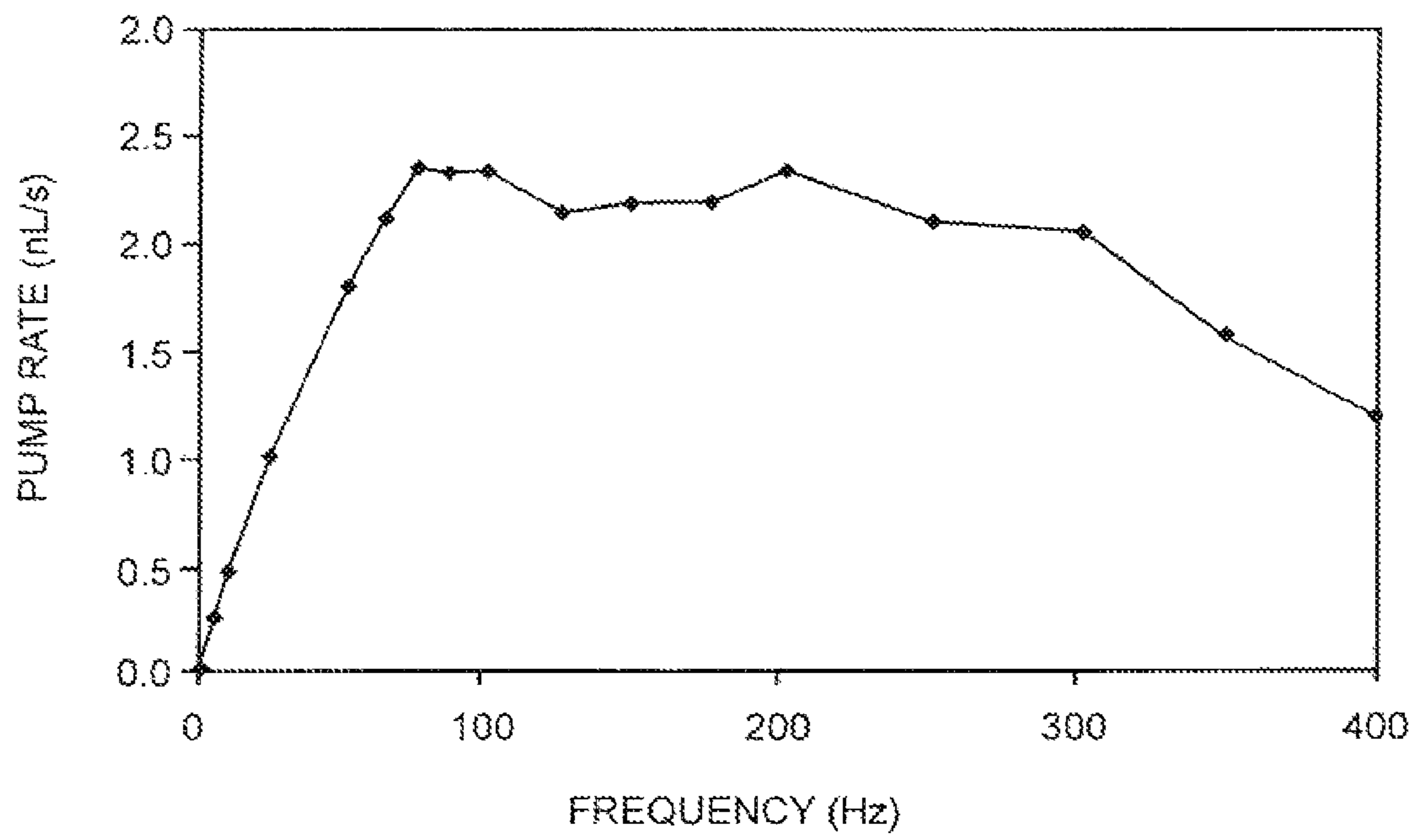


FIG. 14

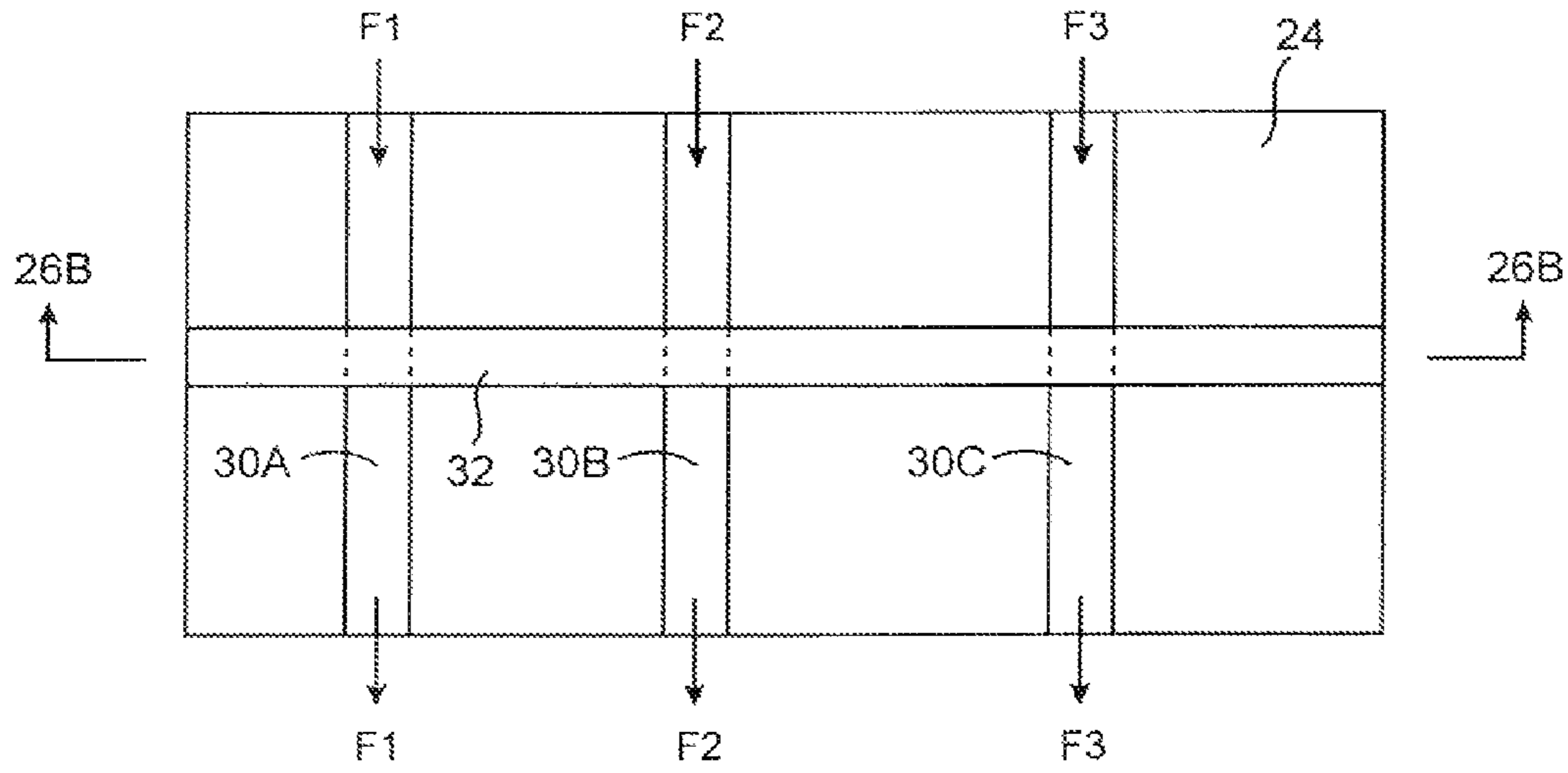


FIG. 15A

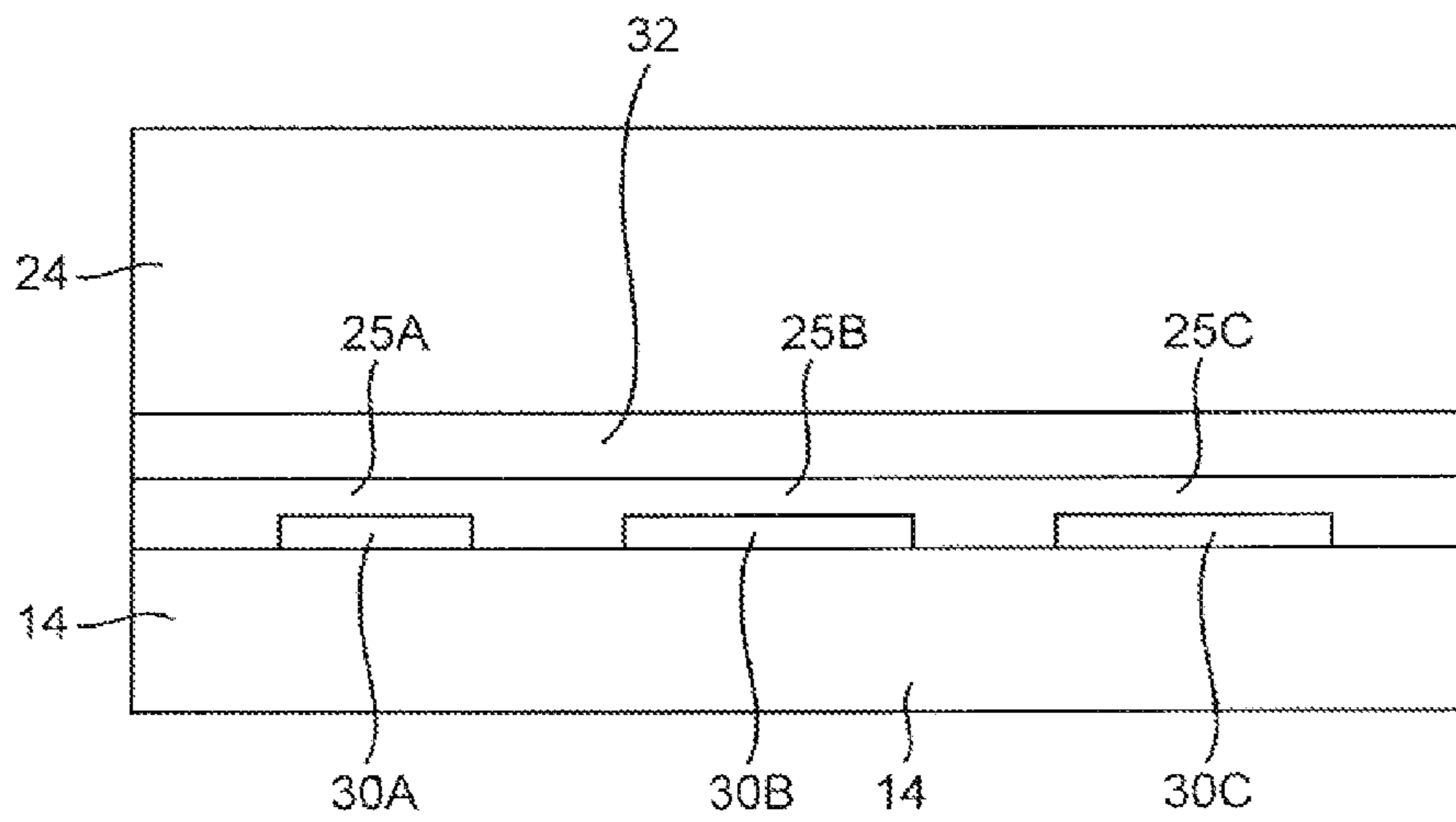


FIG. 15B

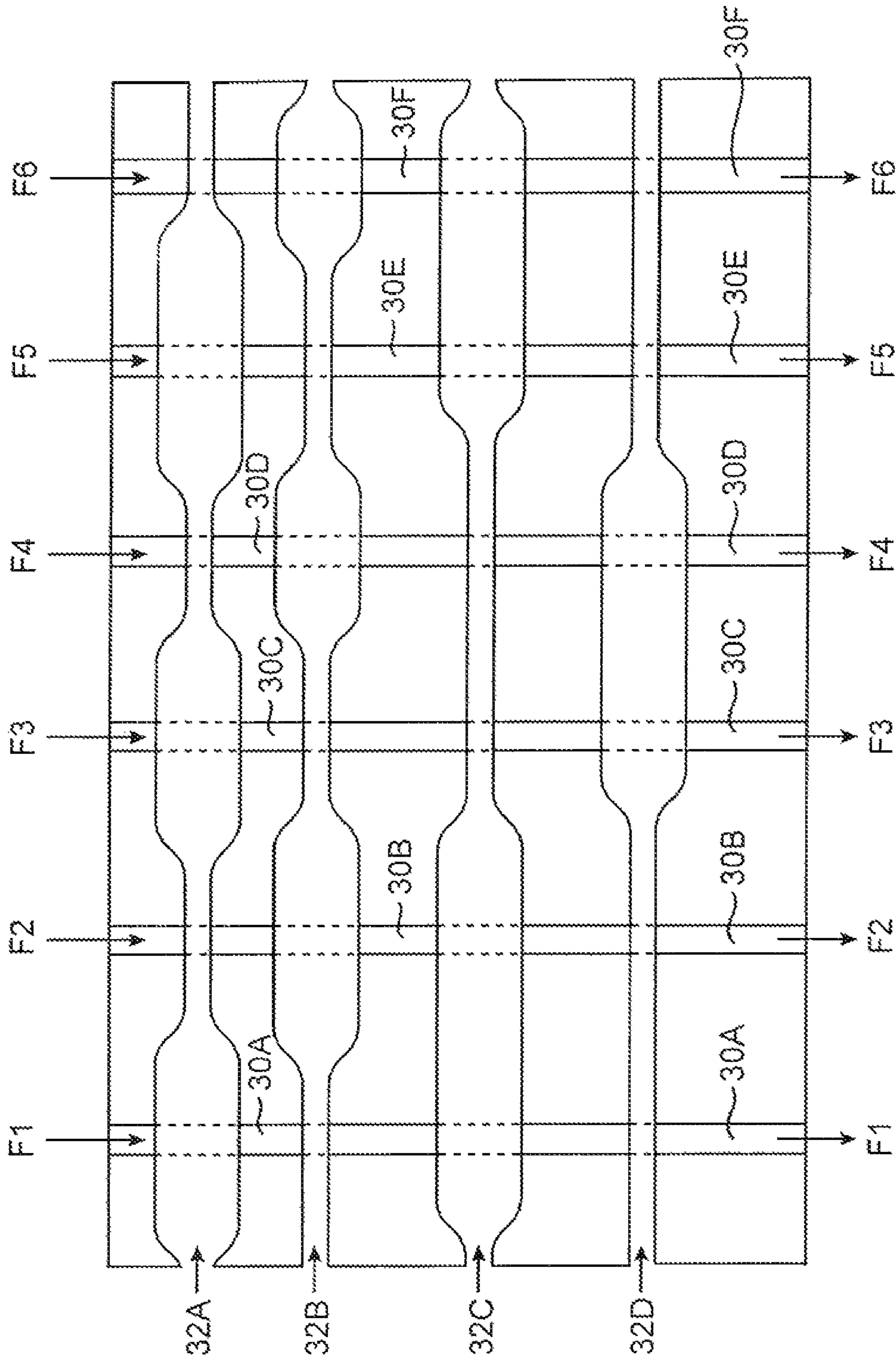


FIG. 16

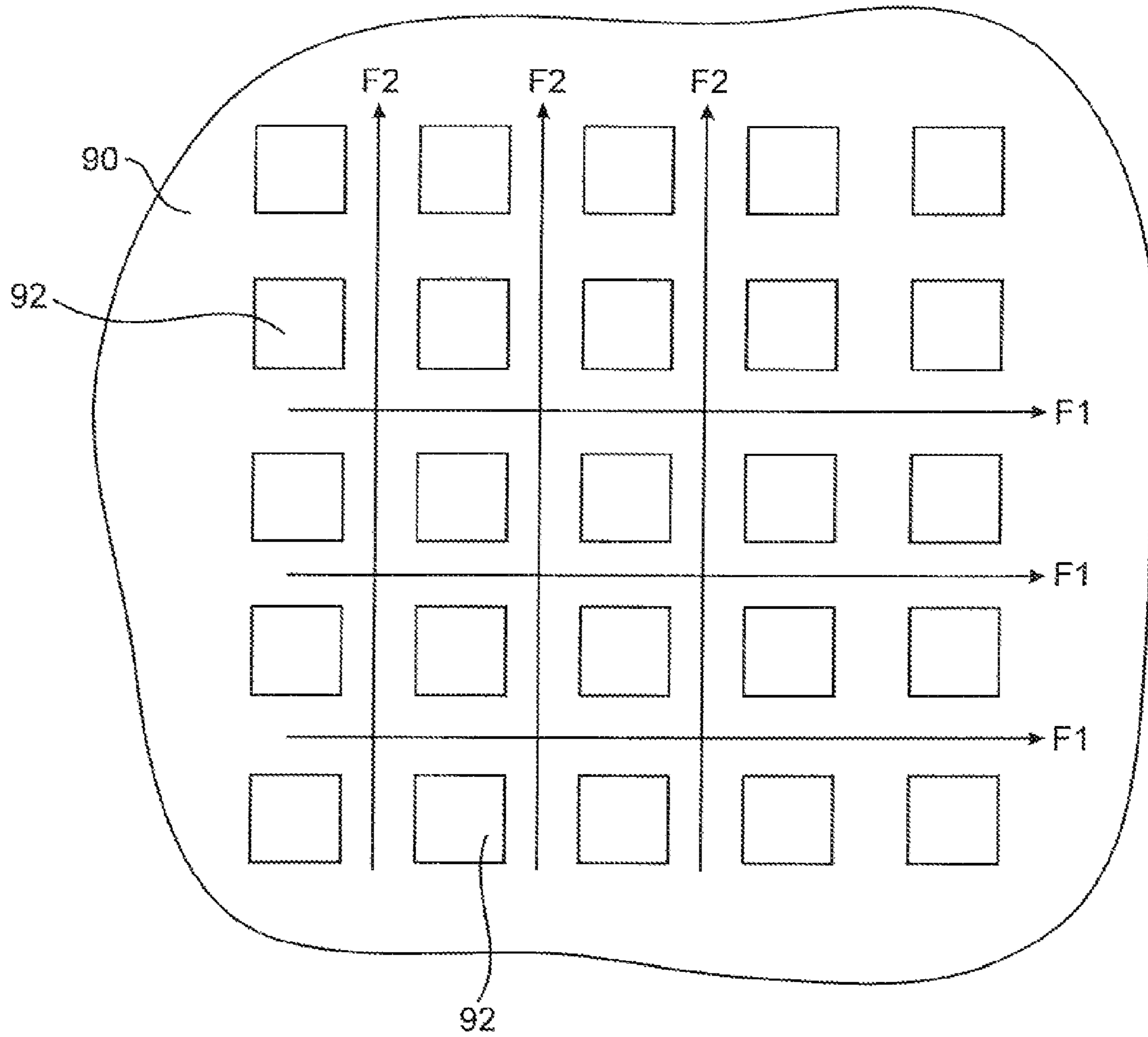


FIG. 17A

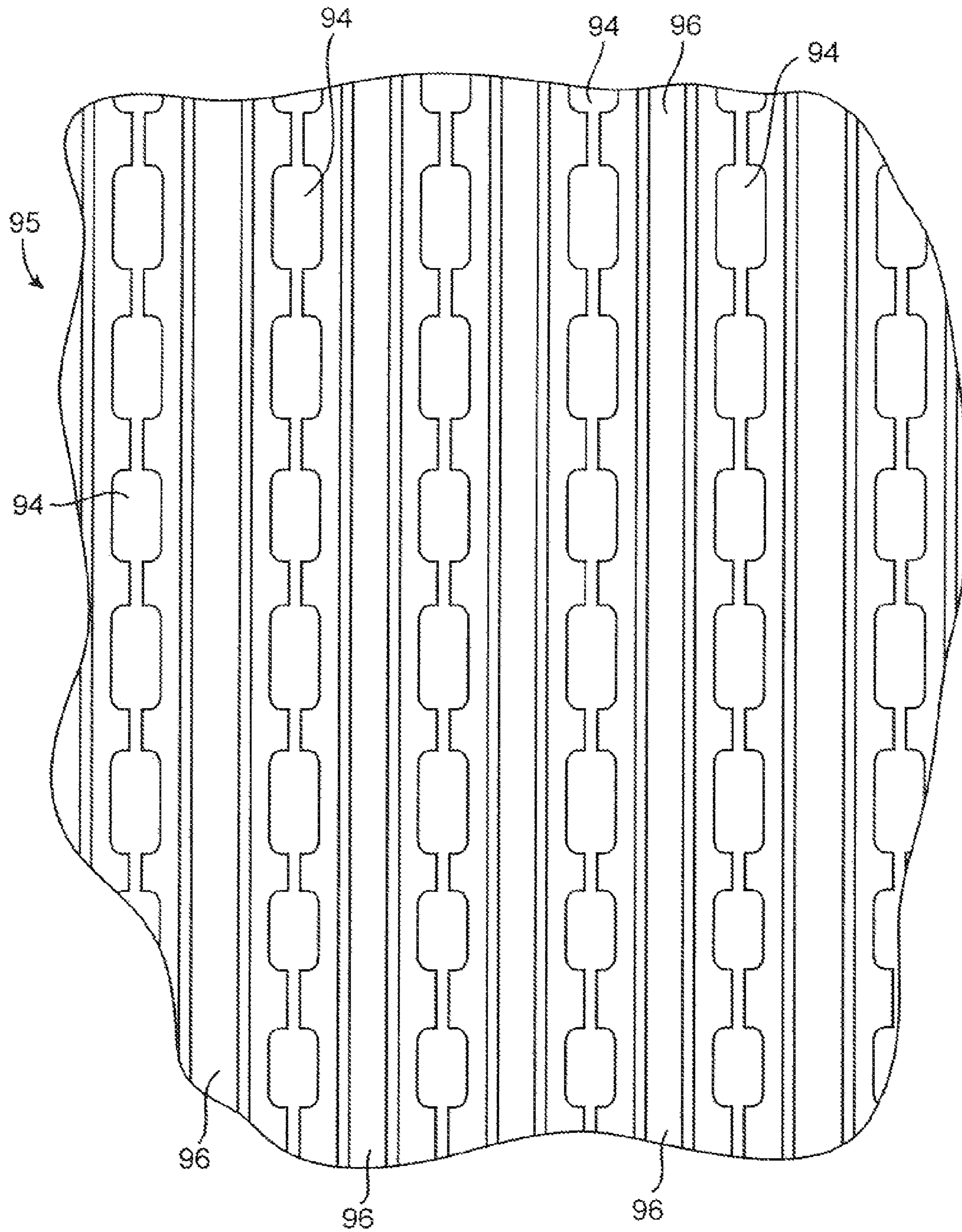


FIG 17B

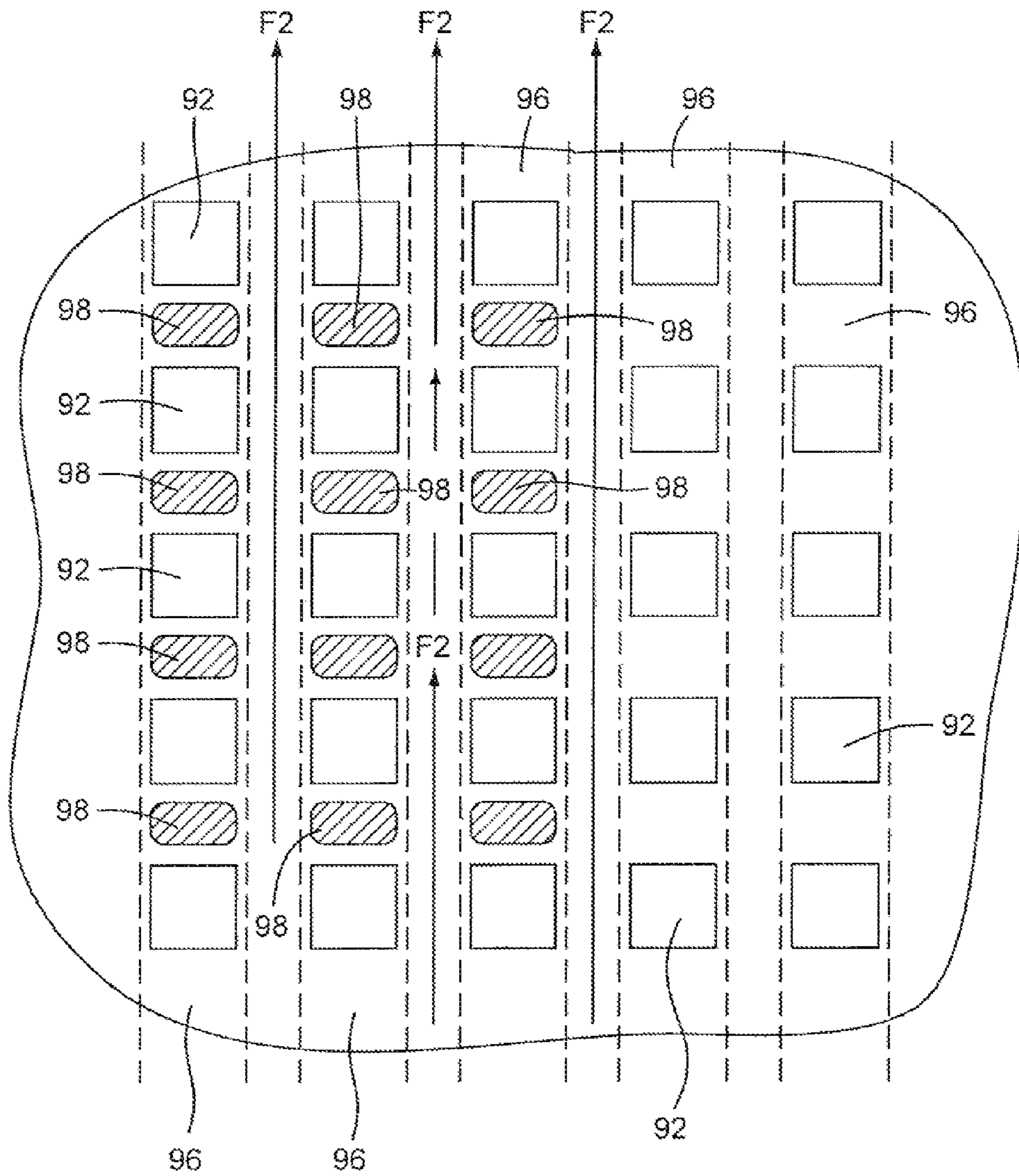


FIG. 17C

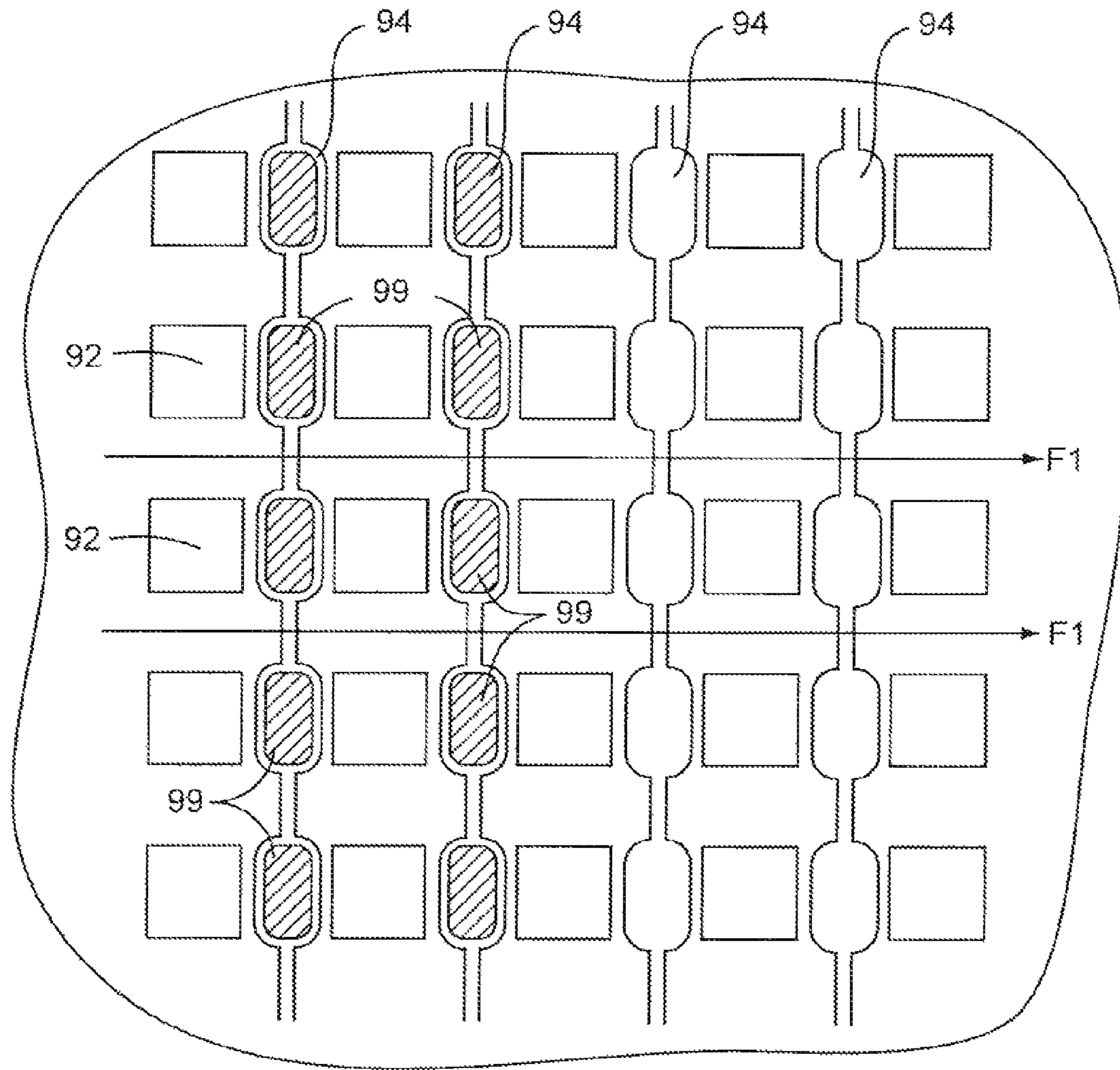


FIG. 17D

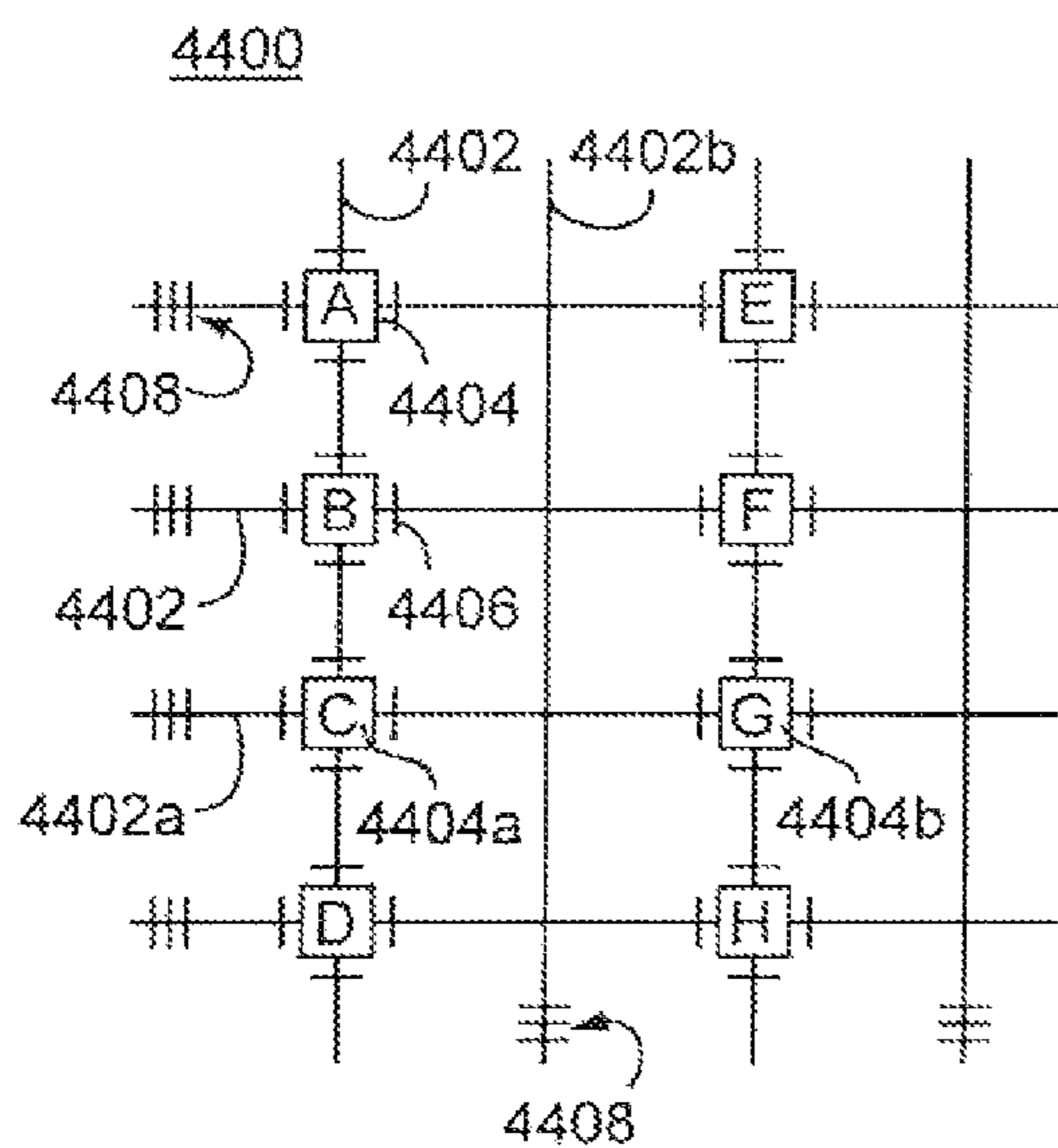


FIG. 18A

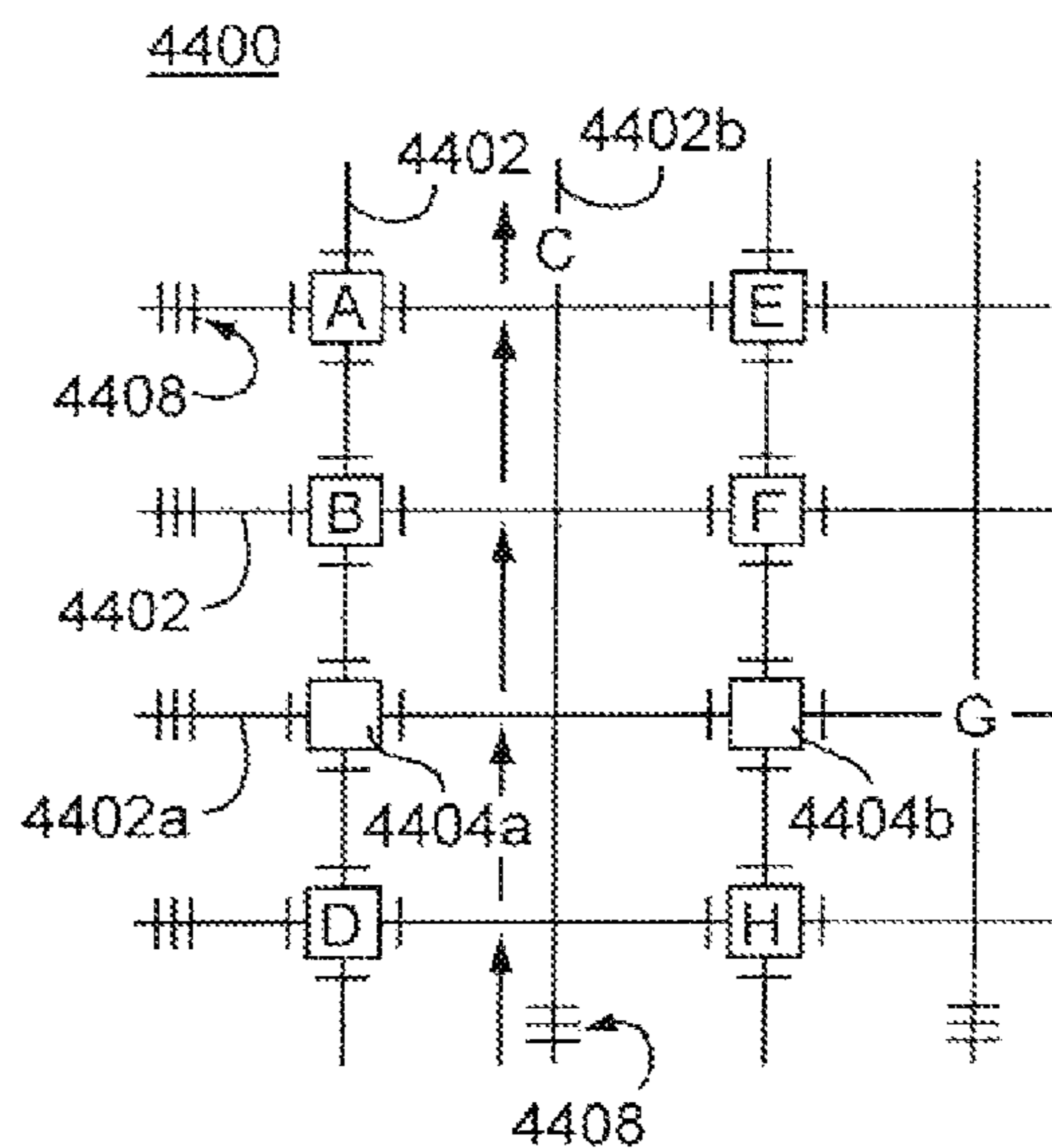


FIG. 18C

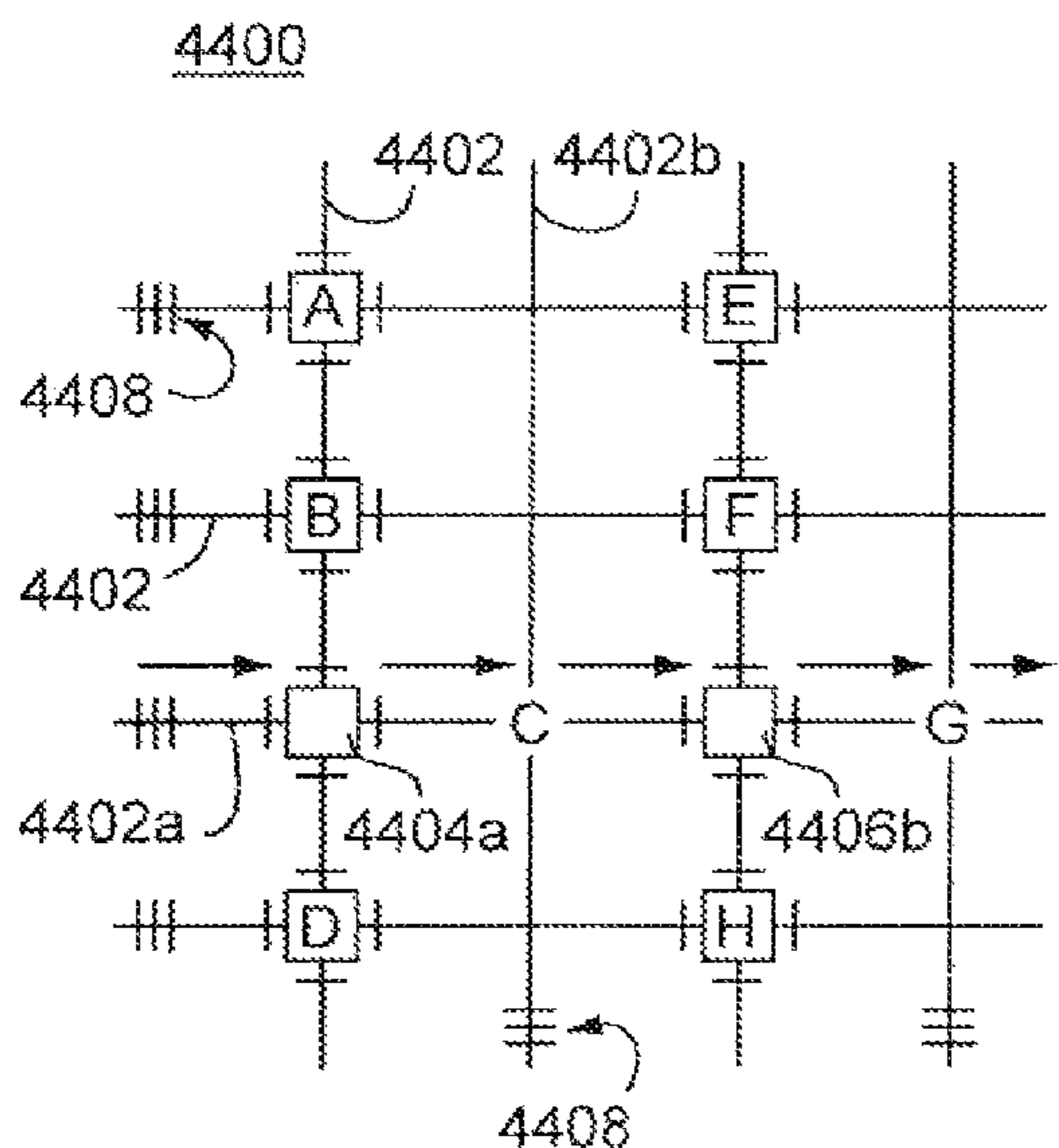


FIG. 18B

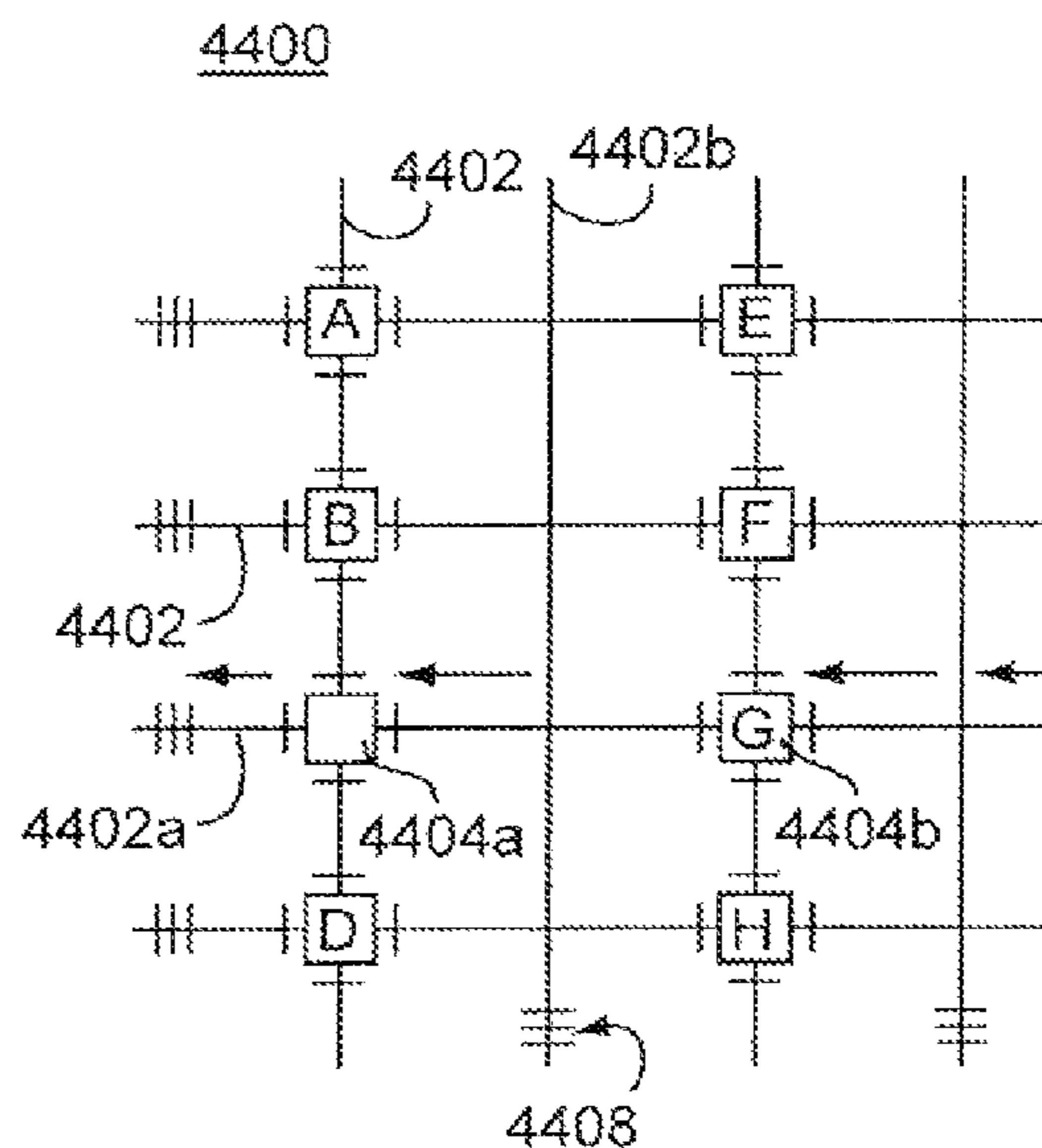


FIG. 18D

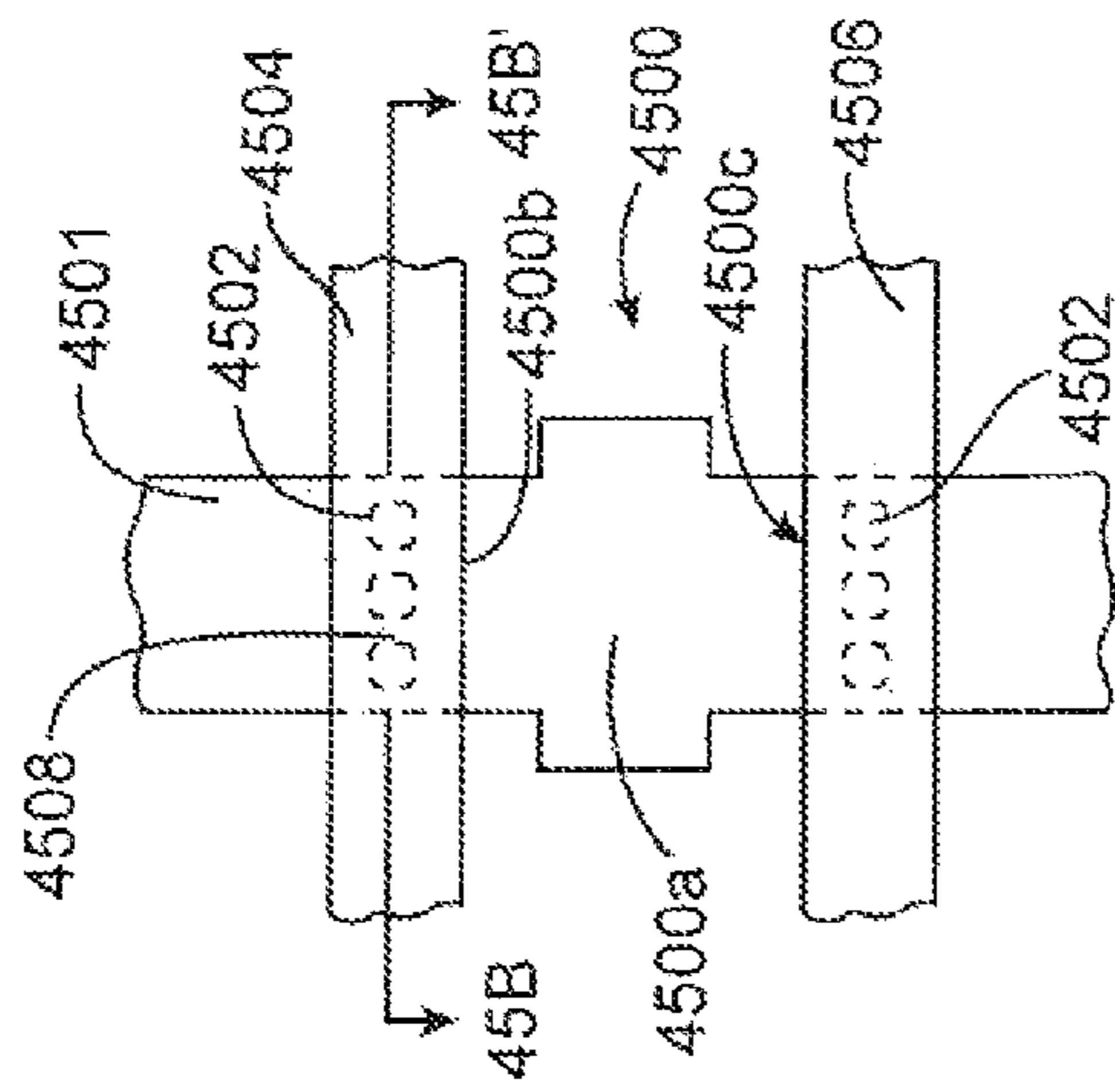


FIG. 19A

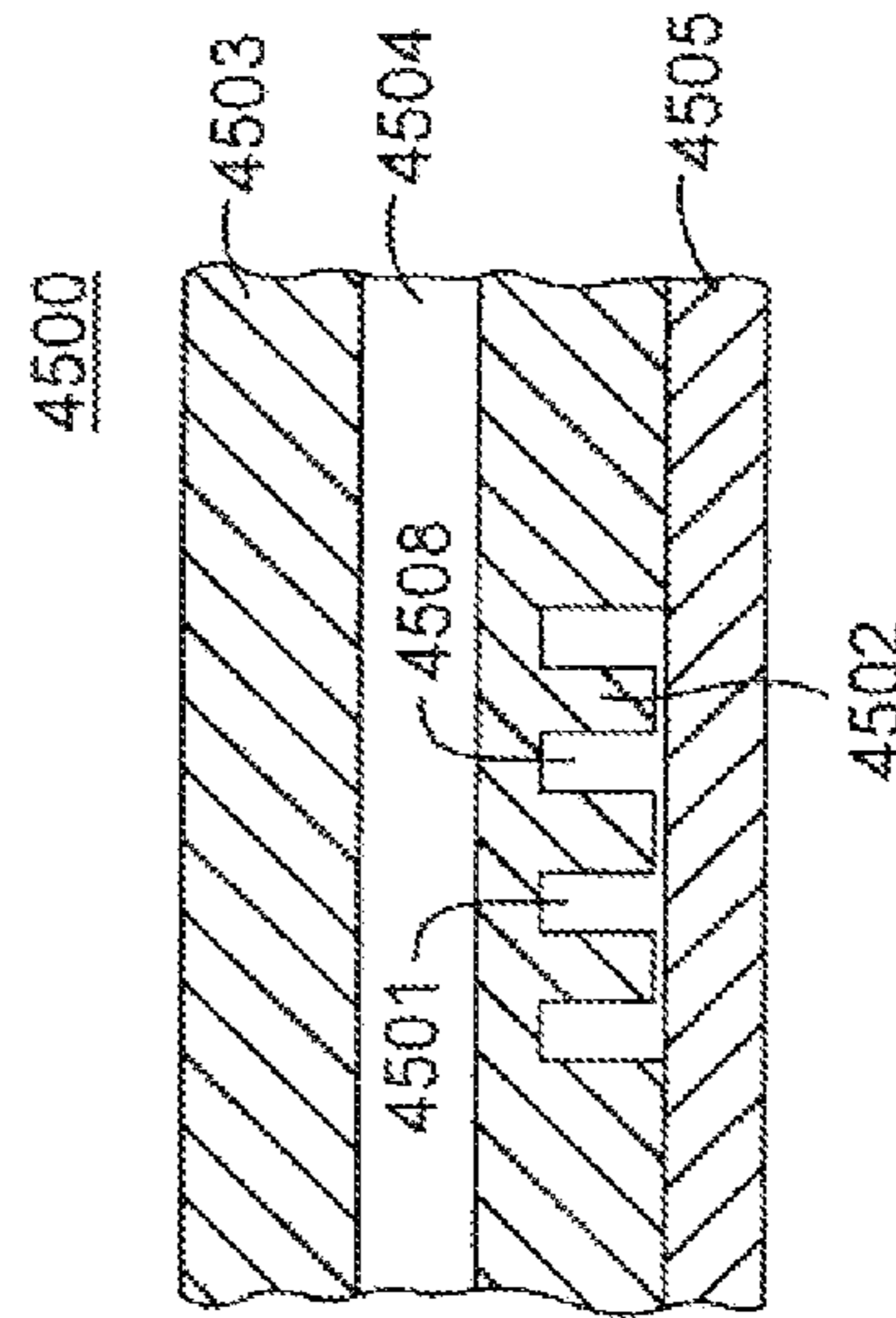


FIG. 19B

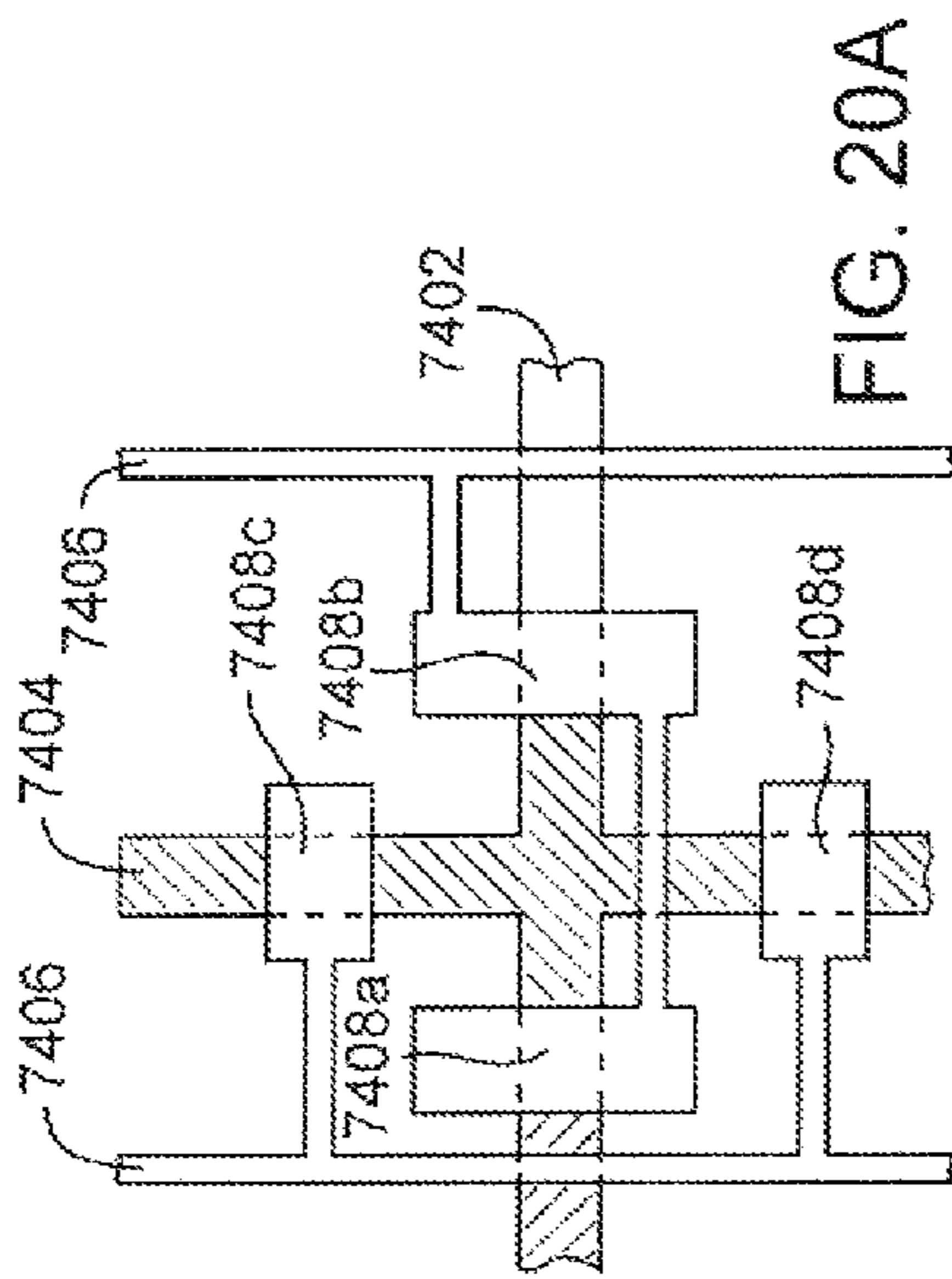


FIG. 20A

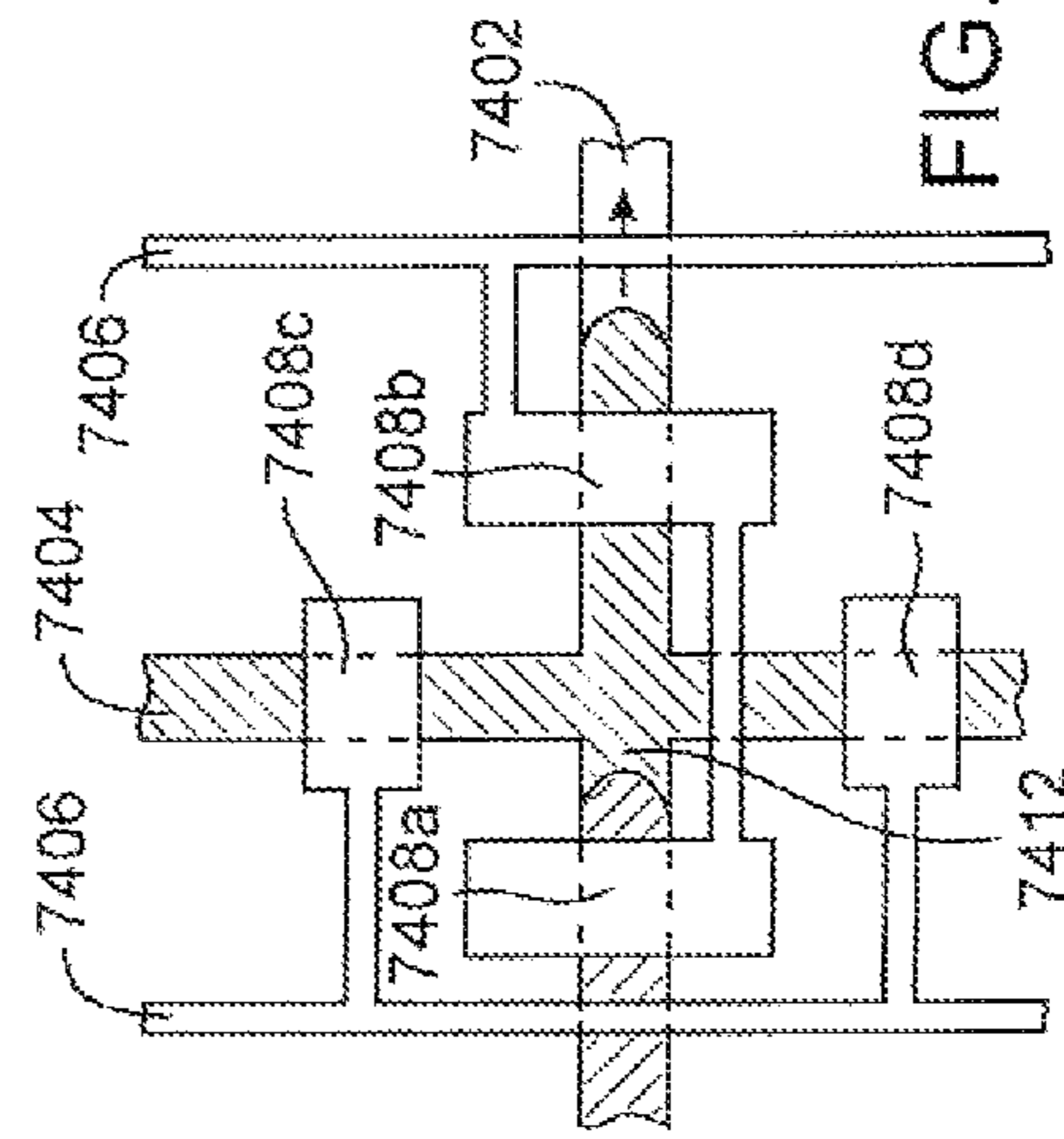


FIG. 20B

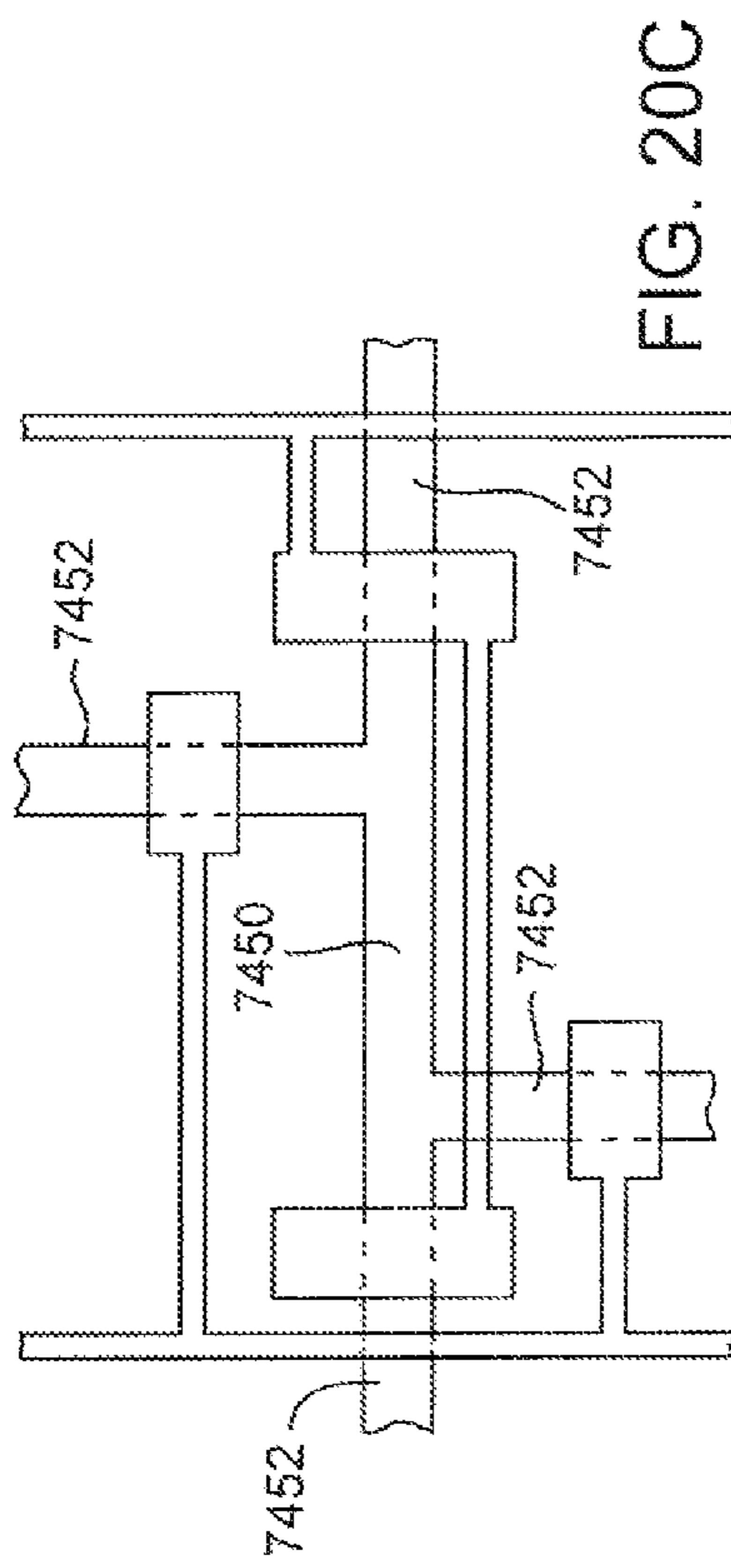


FIG. 20C

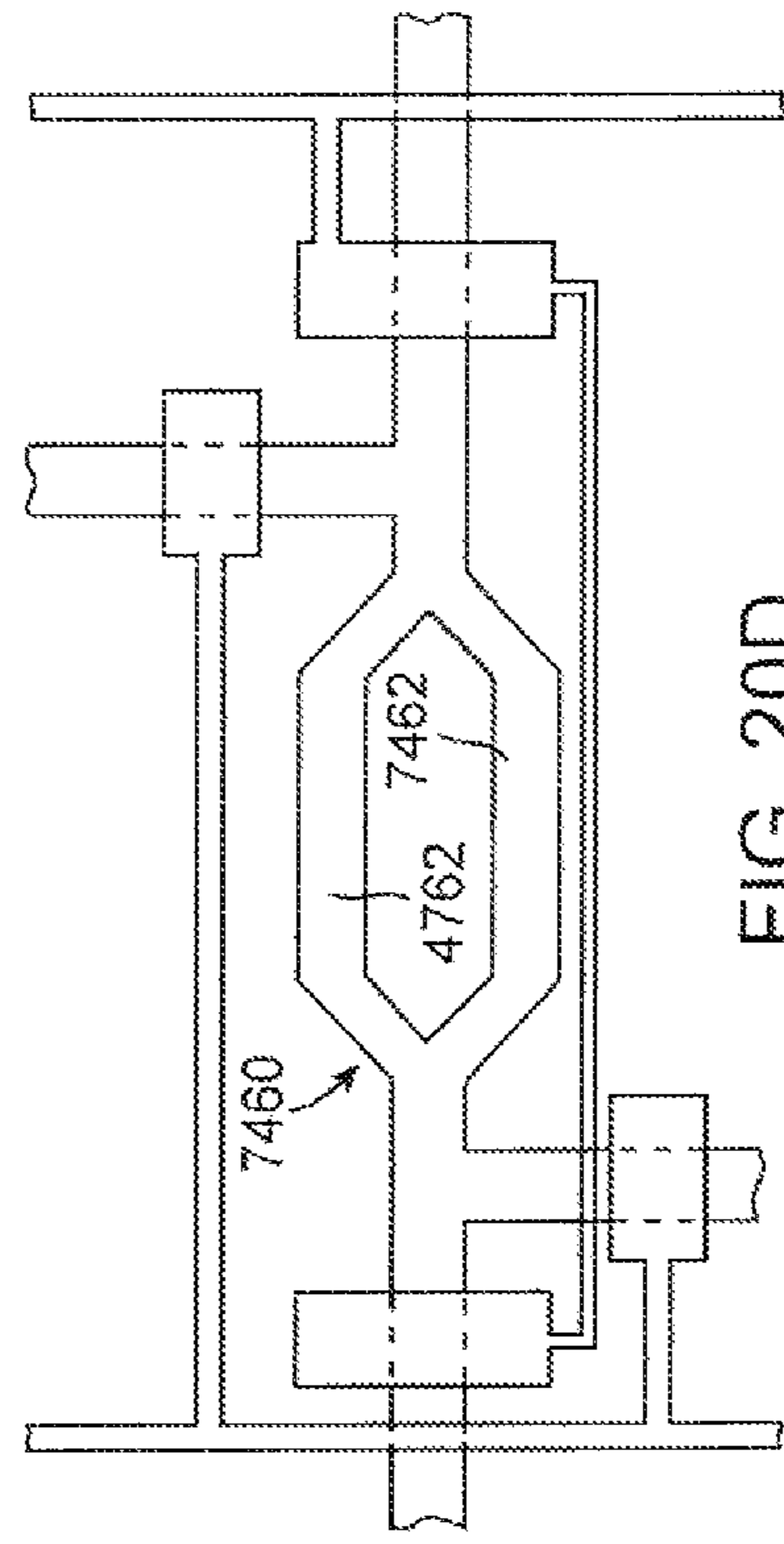


FIG. 20D

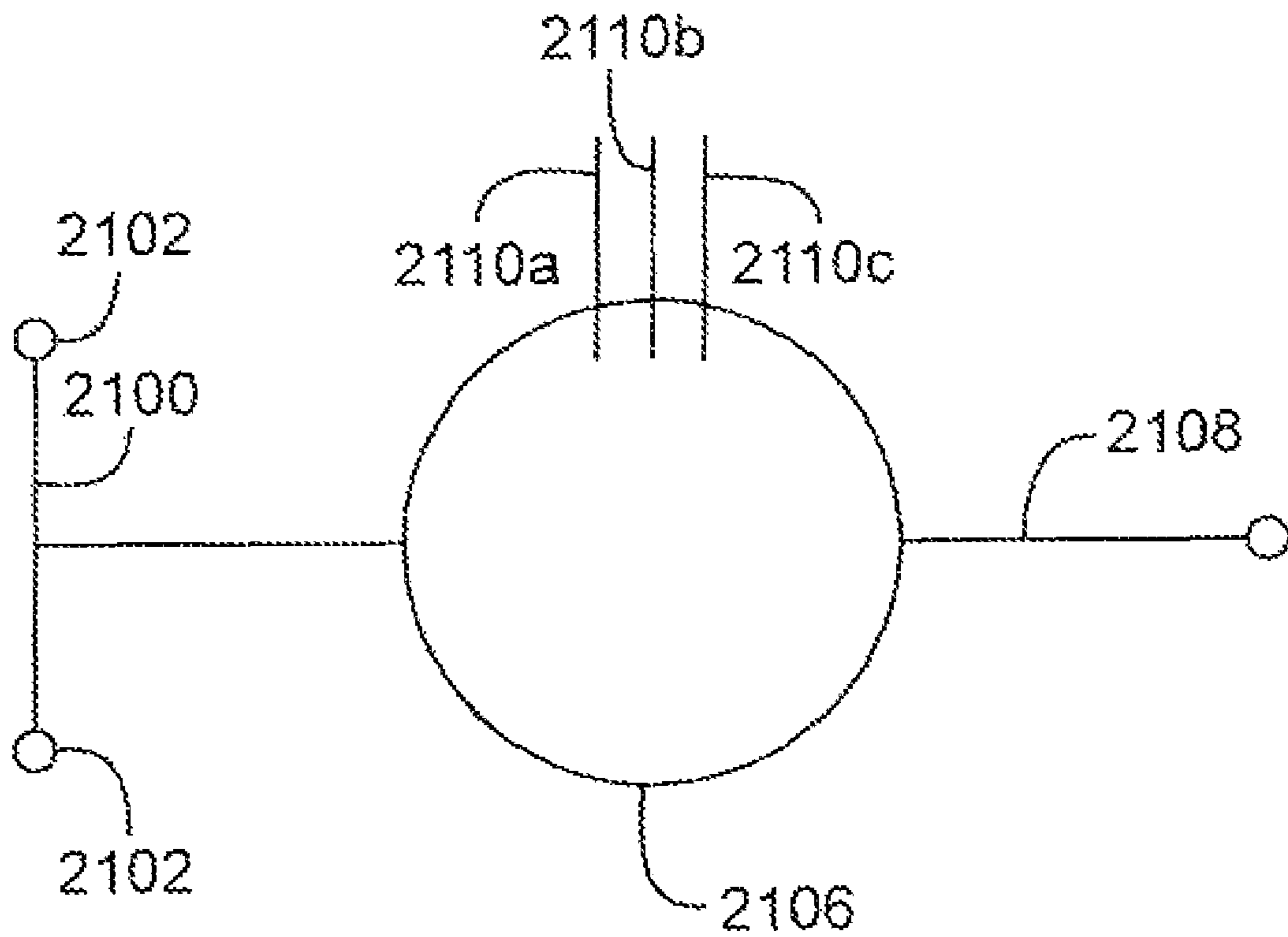


FIG. 21

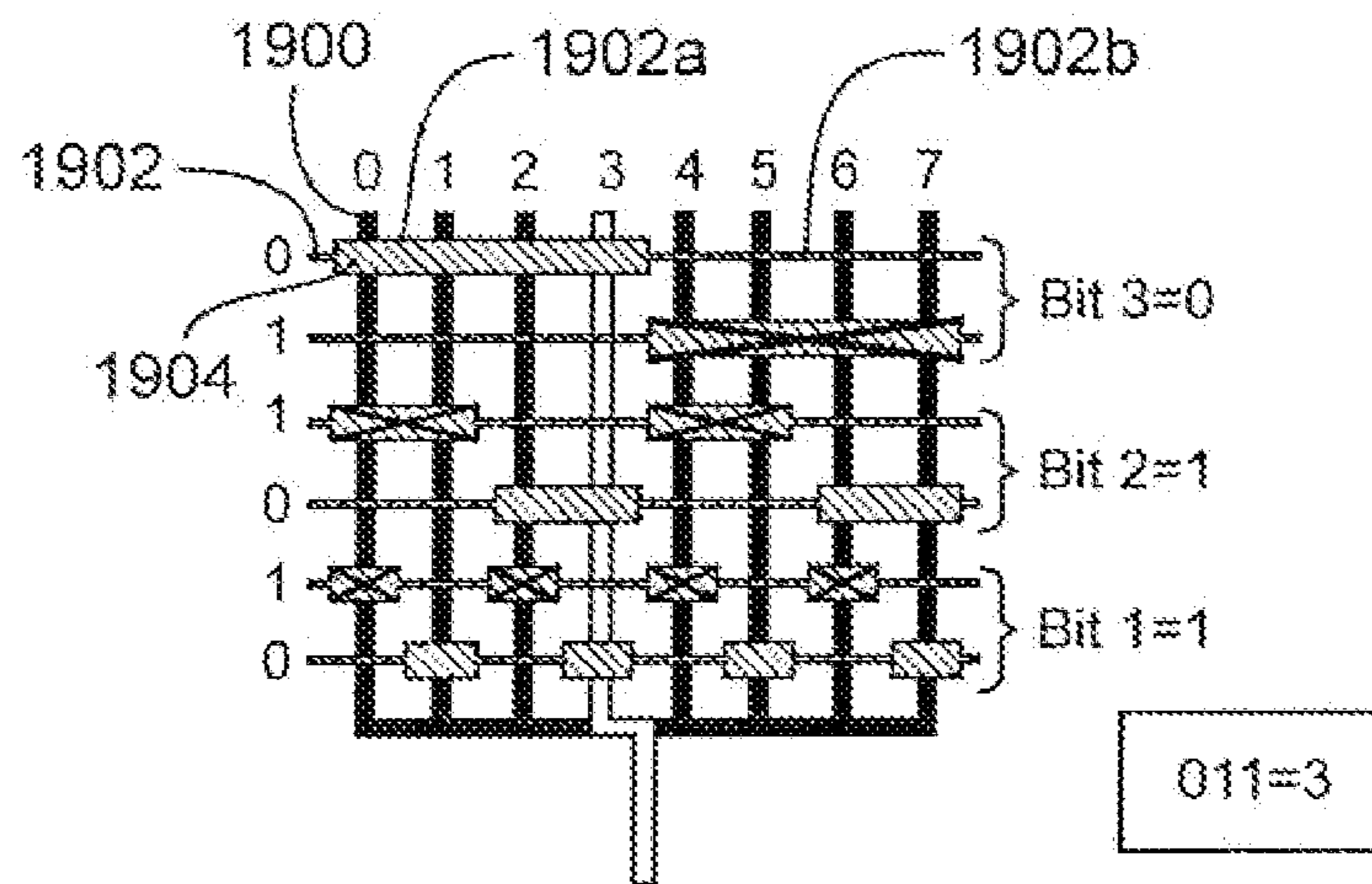


FIG. 22A

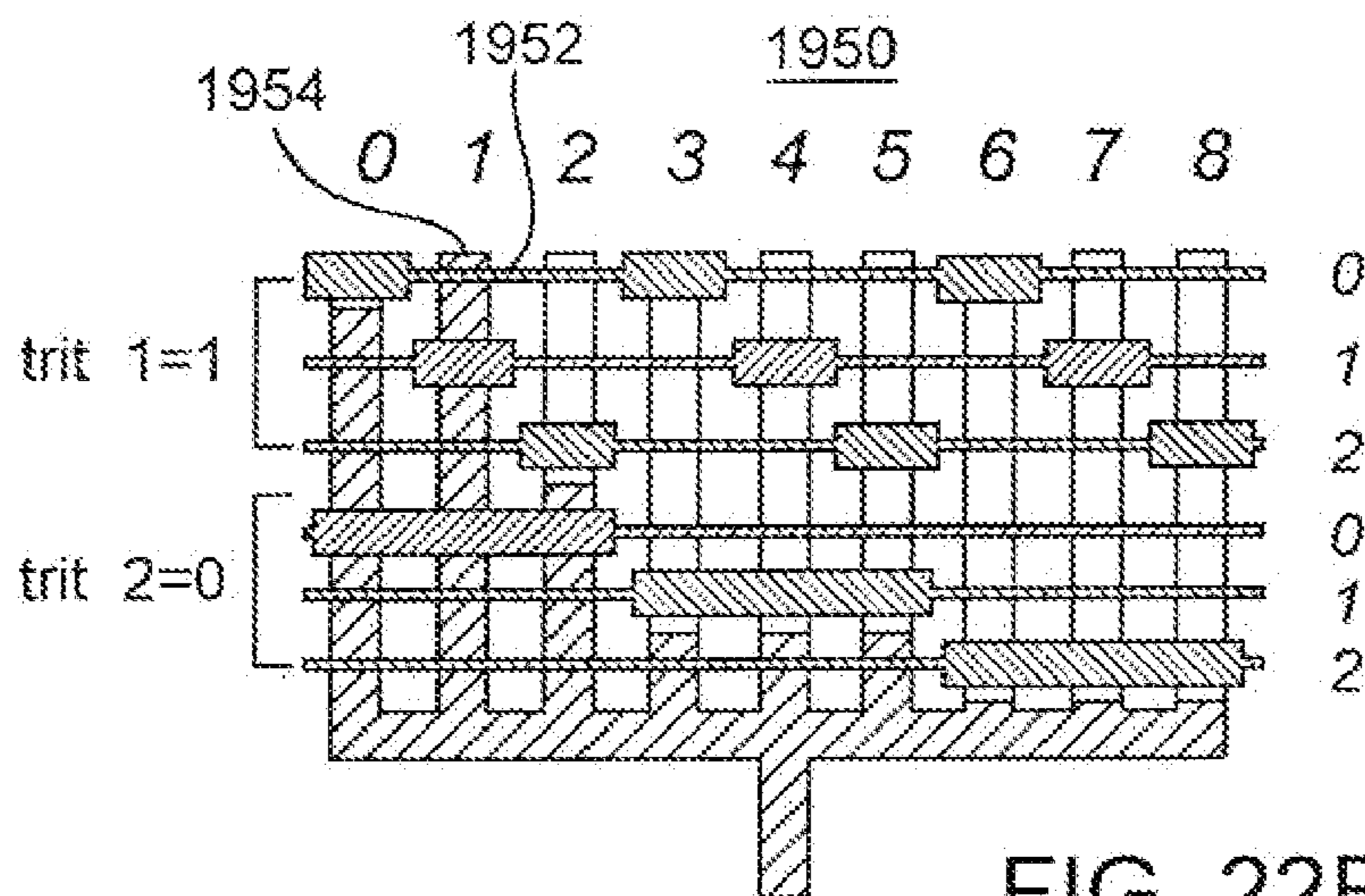


FIG. 22B

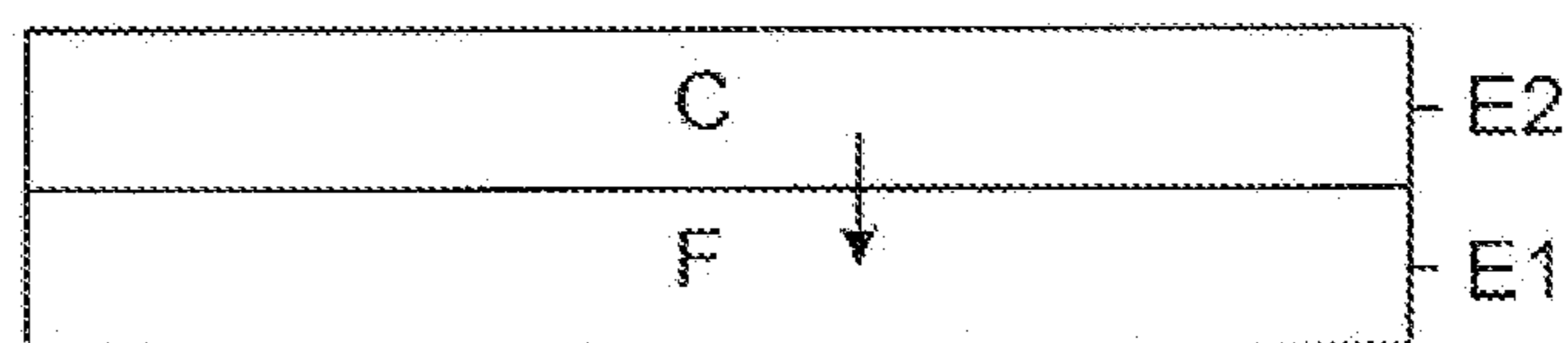
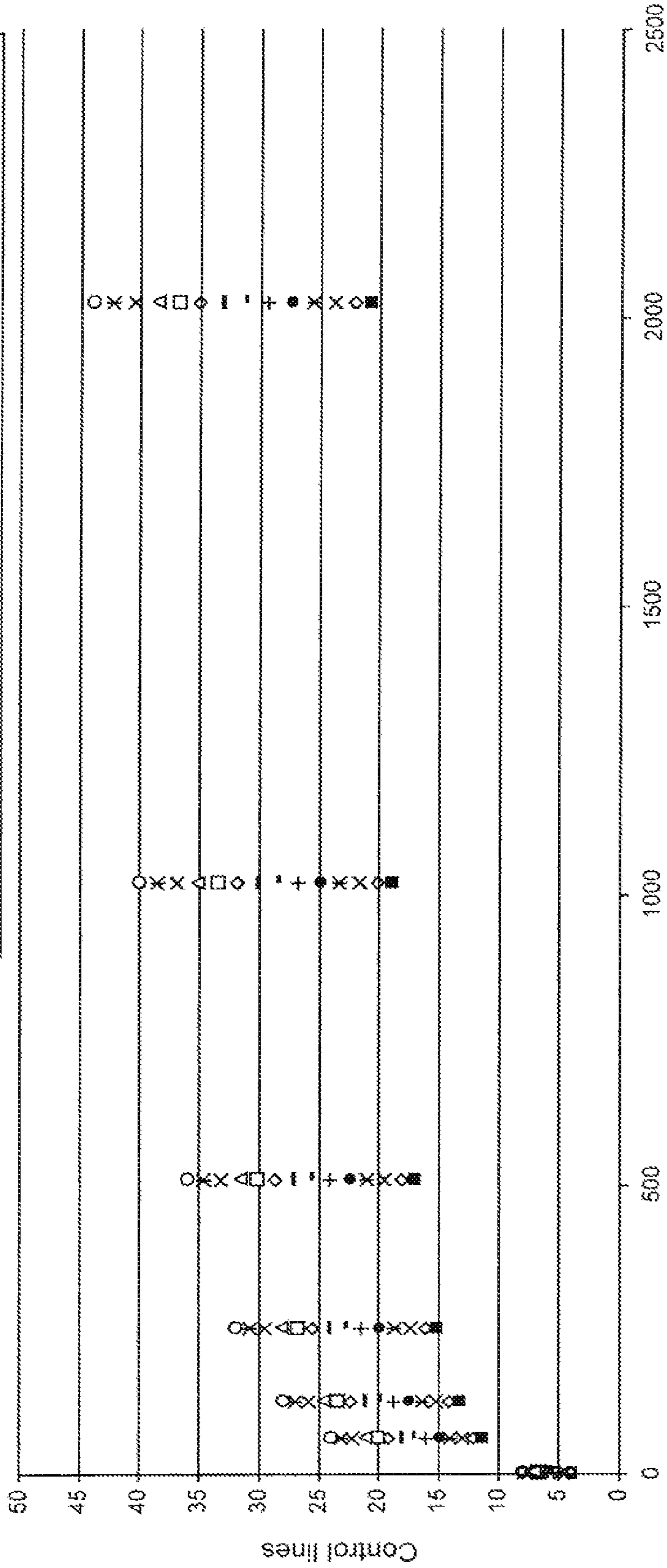


FIG. 22C

Base No. X (Xlogx)

◇ 2 ■ 3 △ 4 × 5 ● 6 + 7 - 8 * 9 = 10 ◇ 11 □ 12 △ 13 × 14 × 15 ○ 16



Flow lines being controlled

FIG. 23

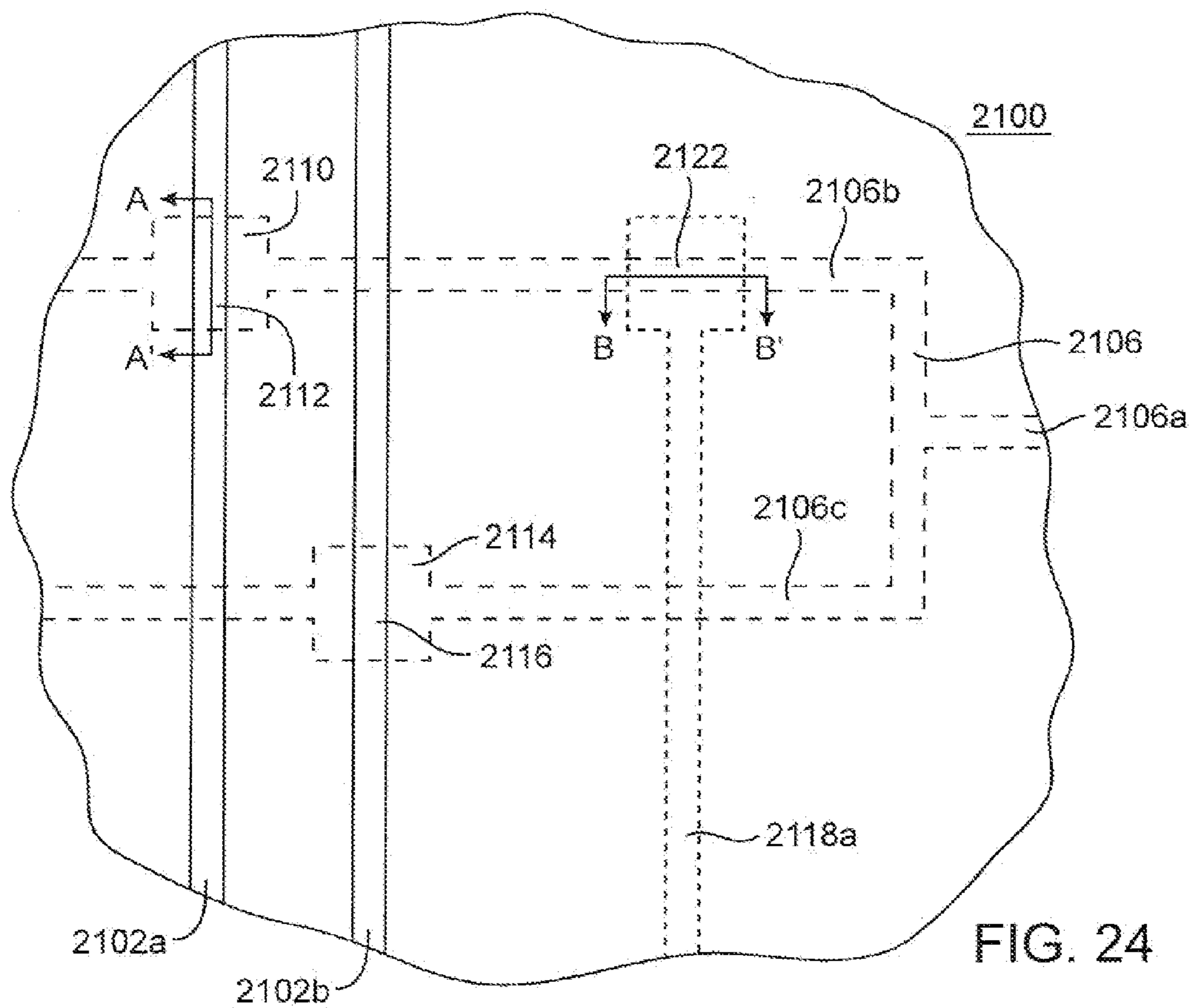


FIG. 24

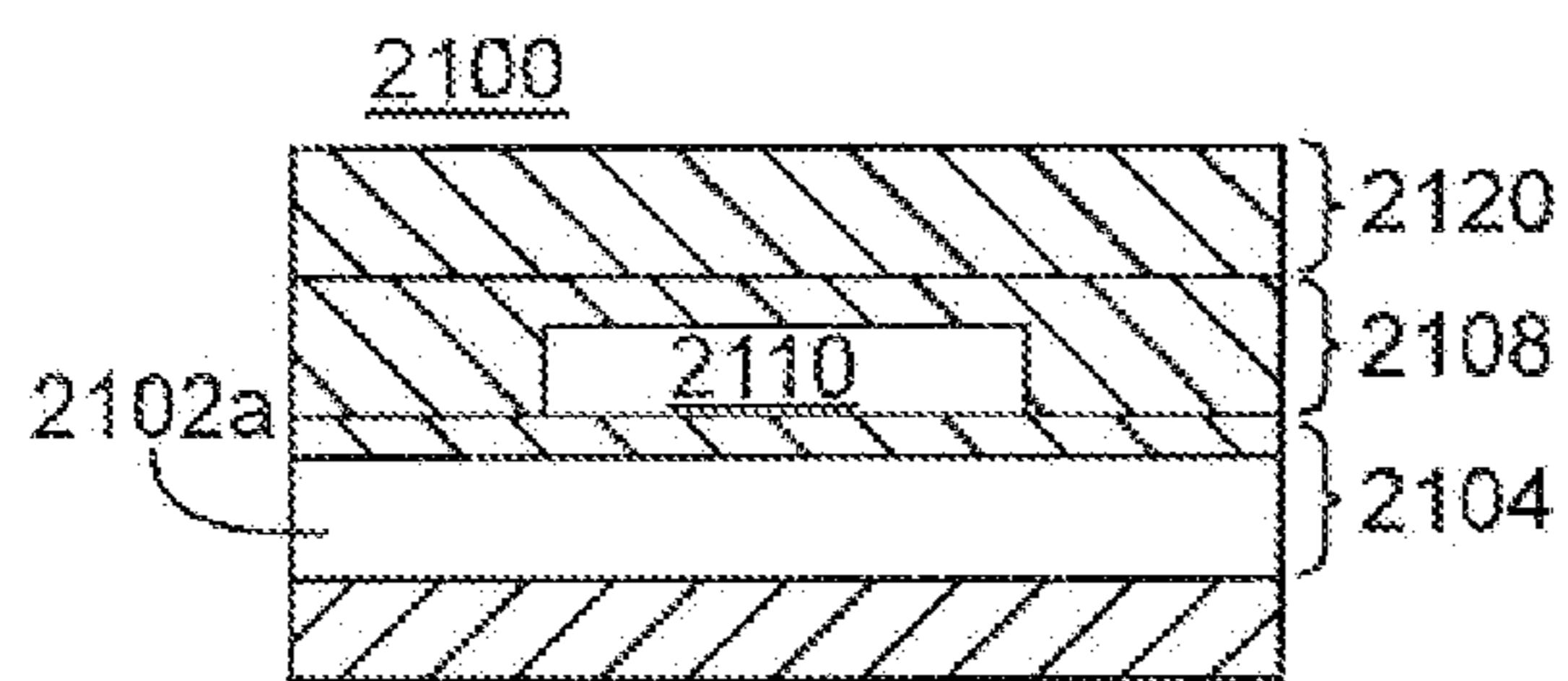


FIG. 24A

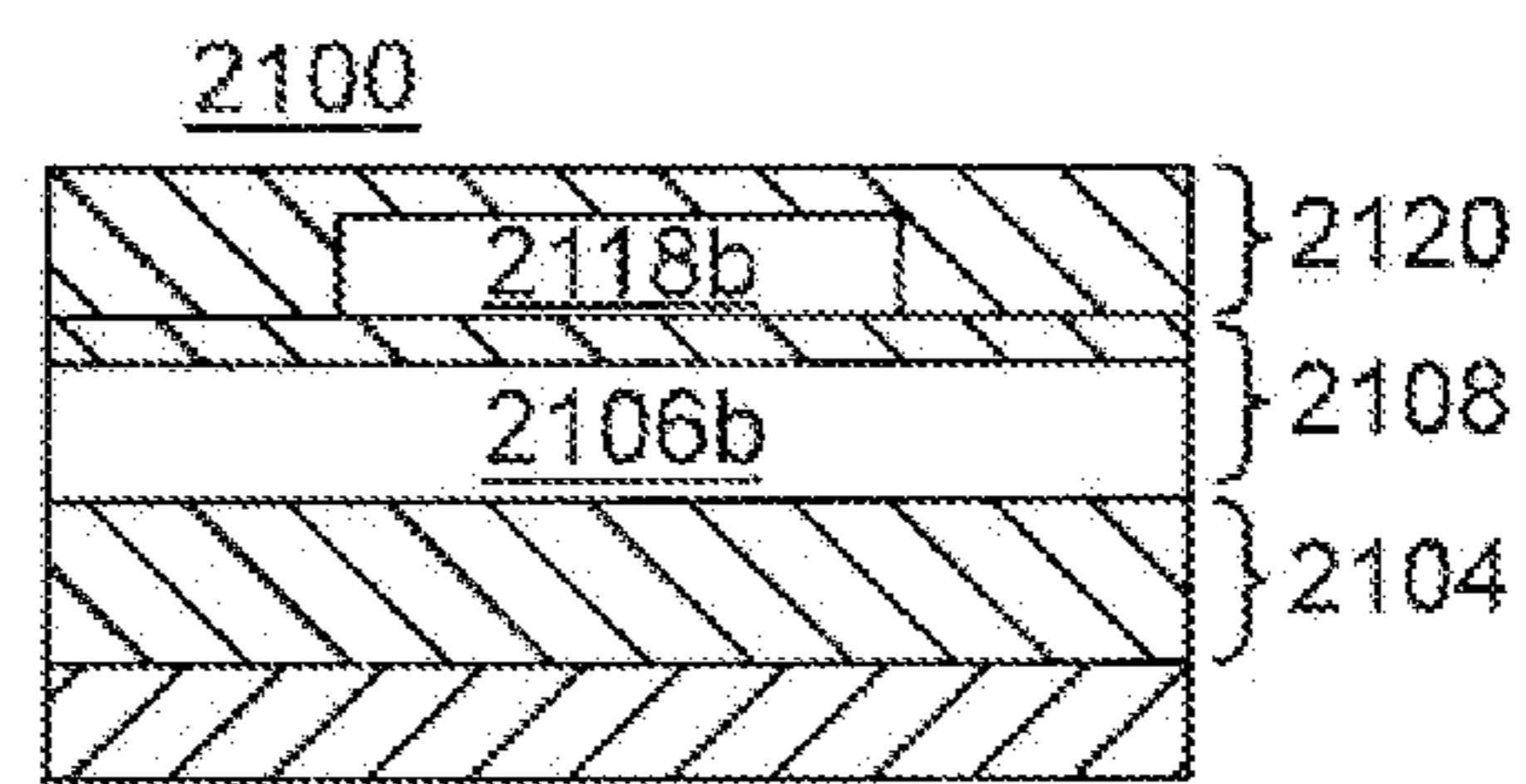


FIG. 24B

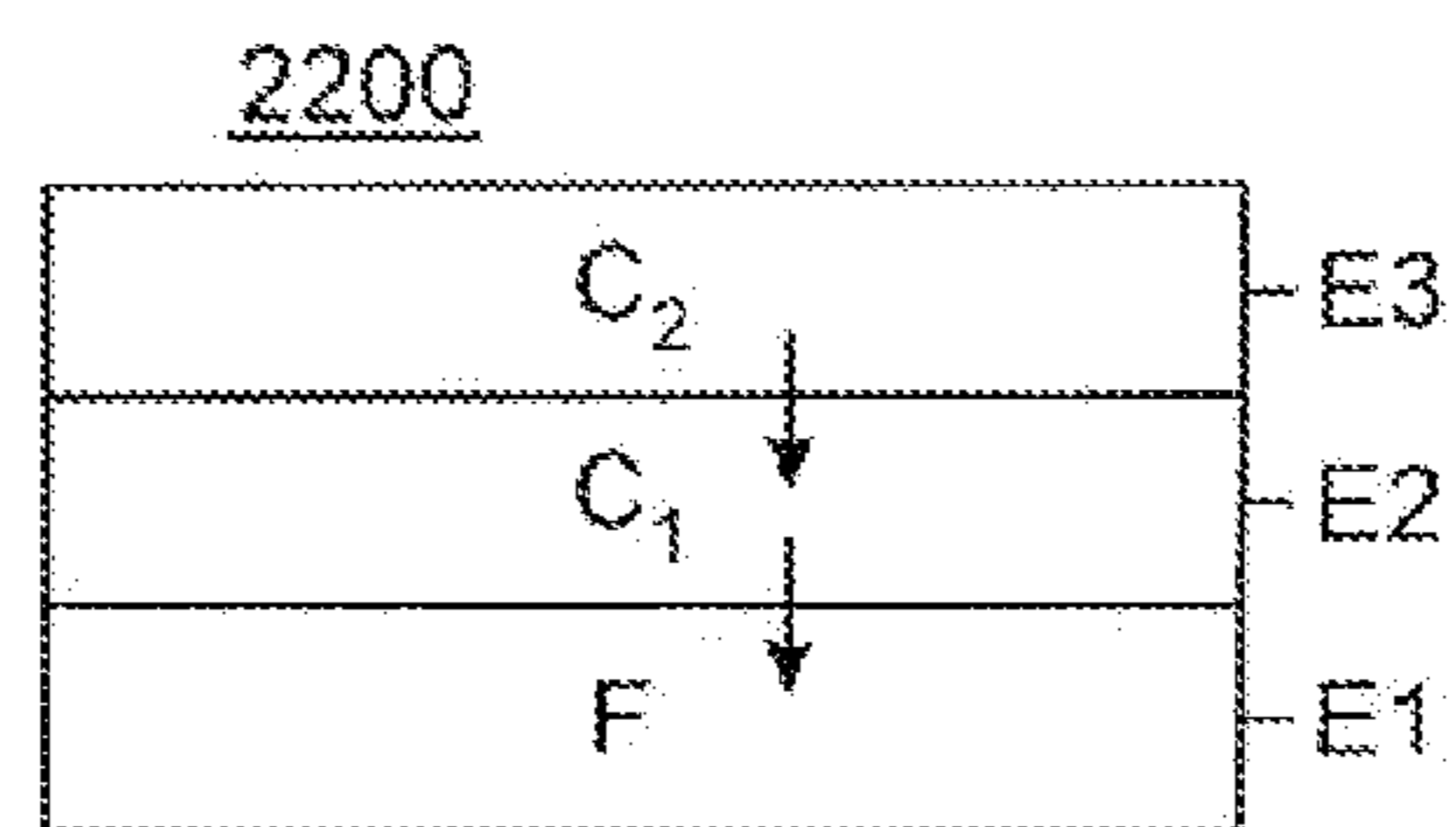


FIG. 25

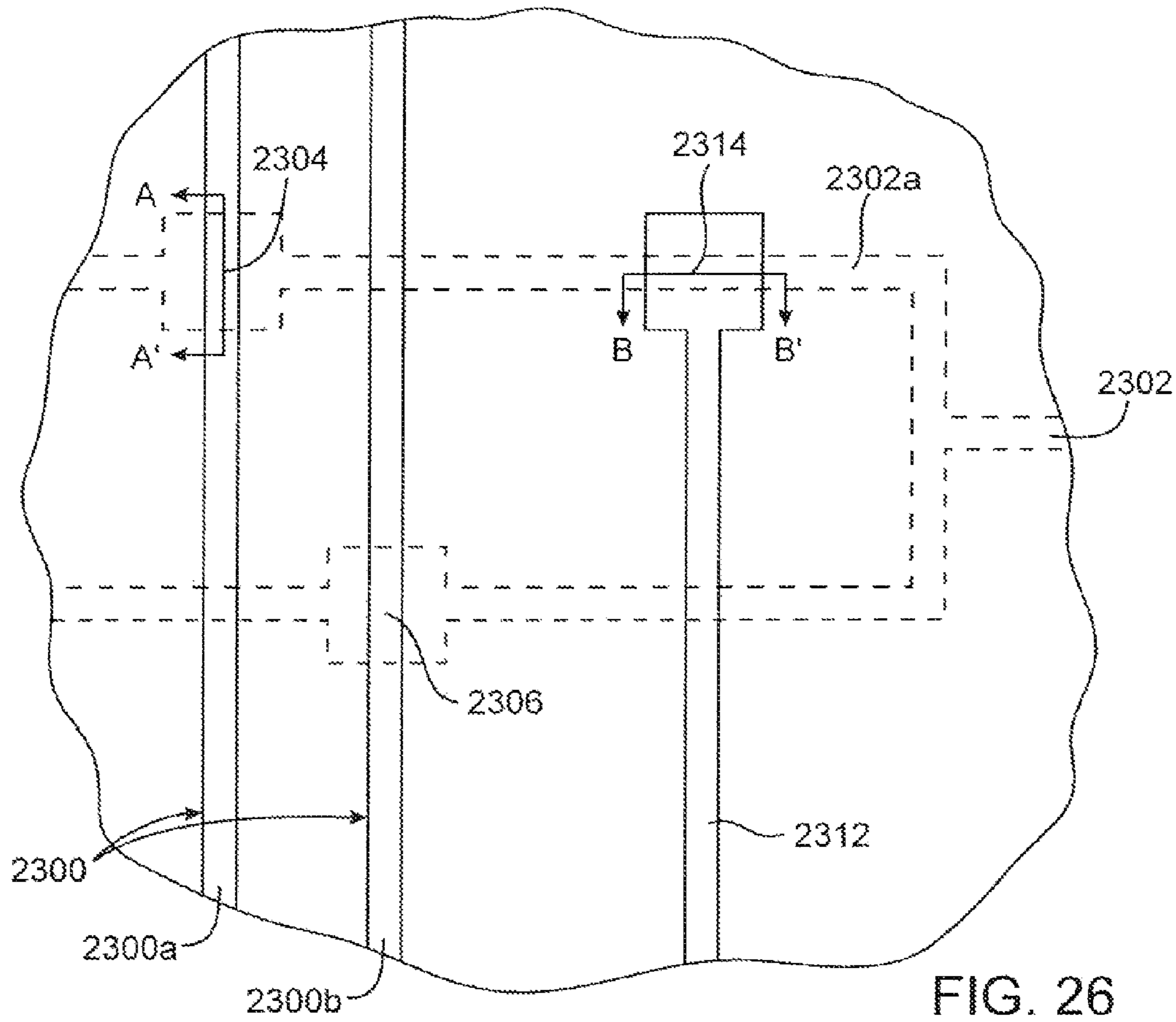


FIG. 26

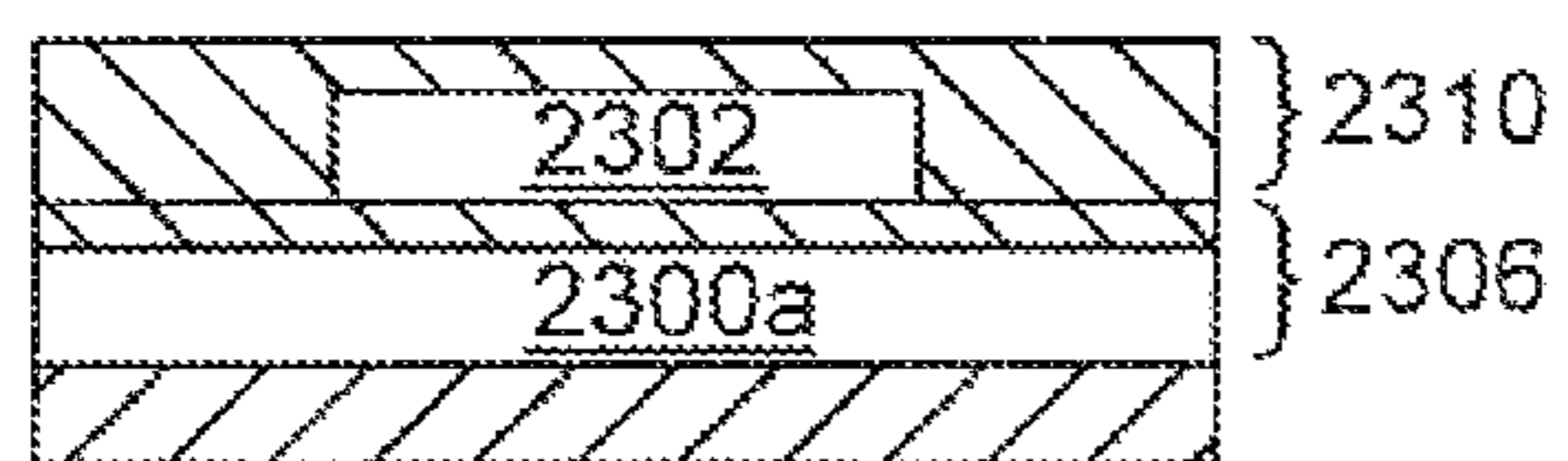


FIG. 26A

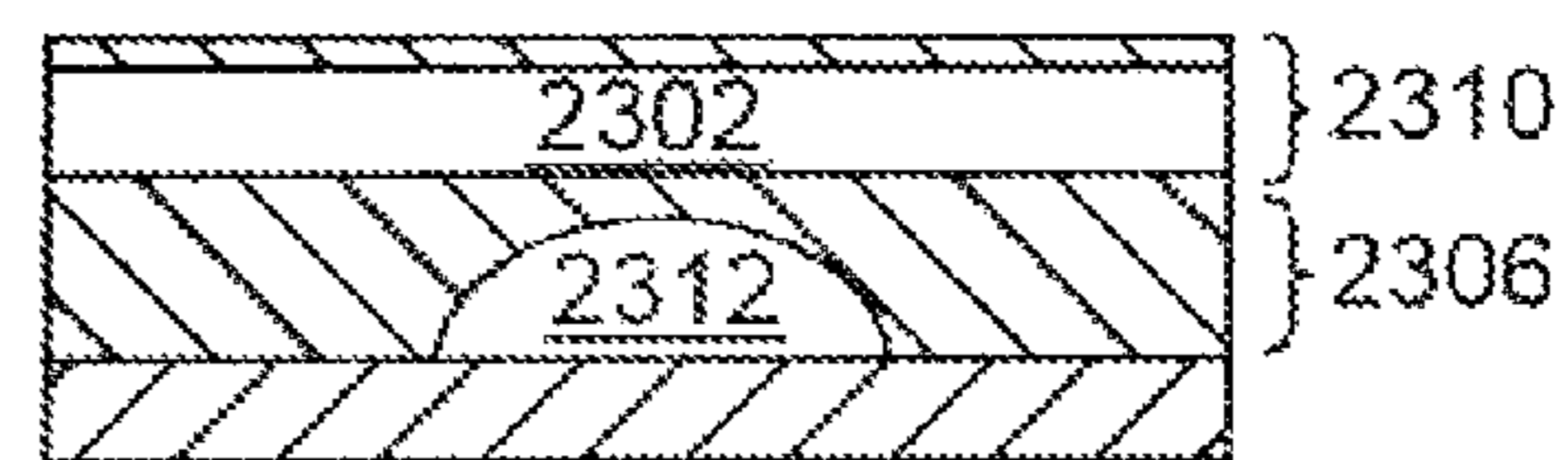


FIG. 26B

3000

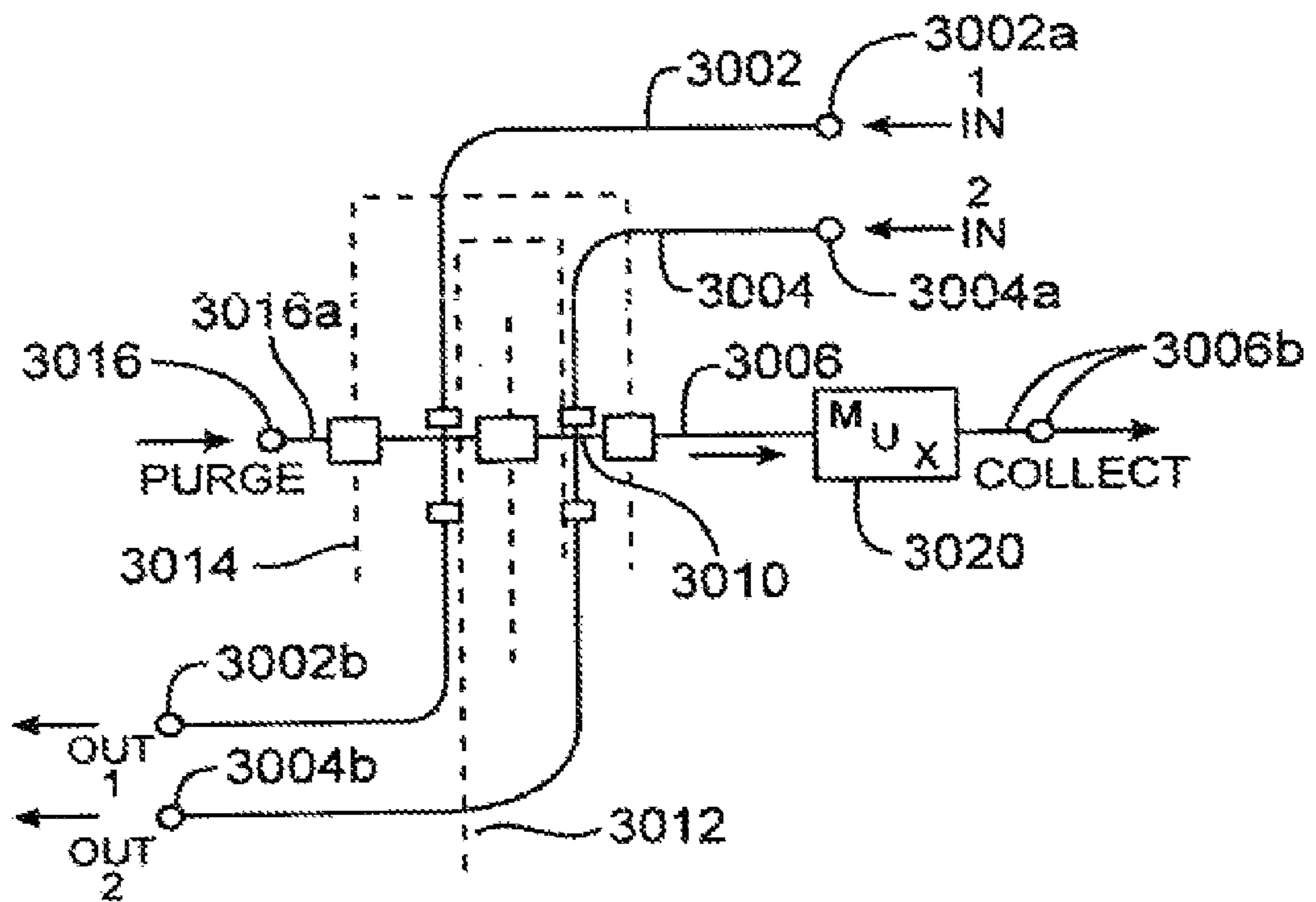


FIG. 27A

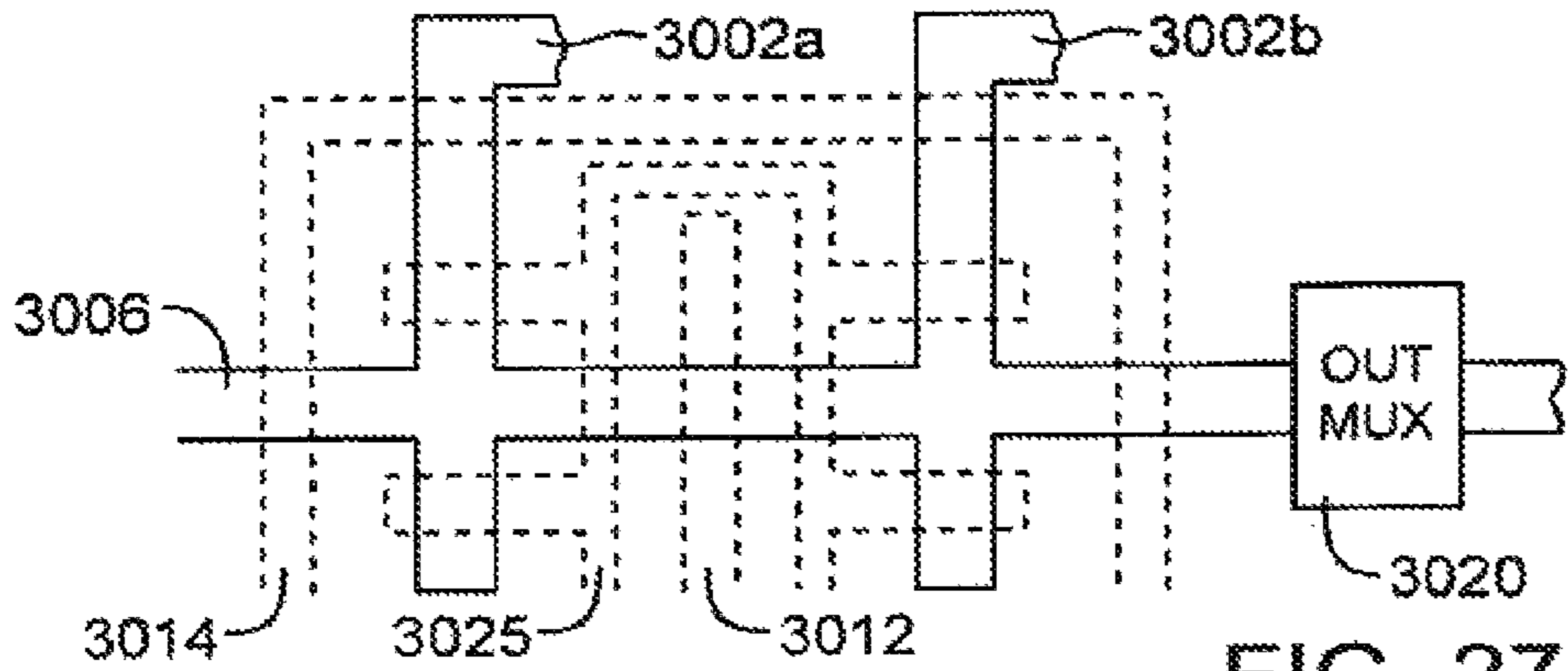


FIG. 27B

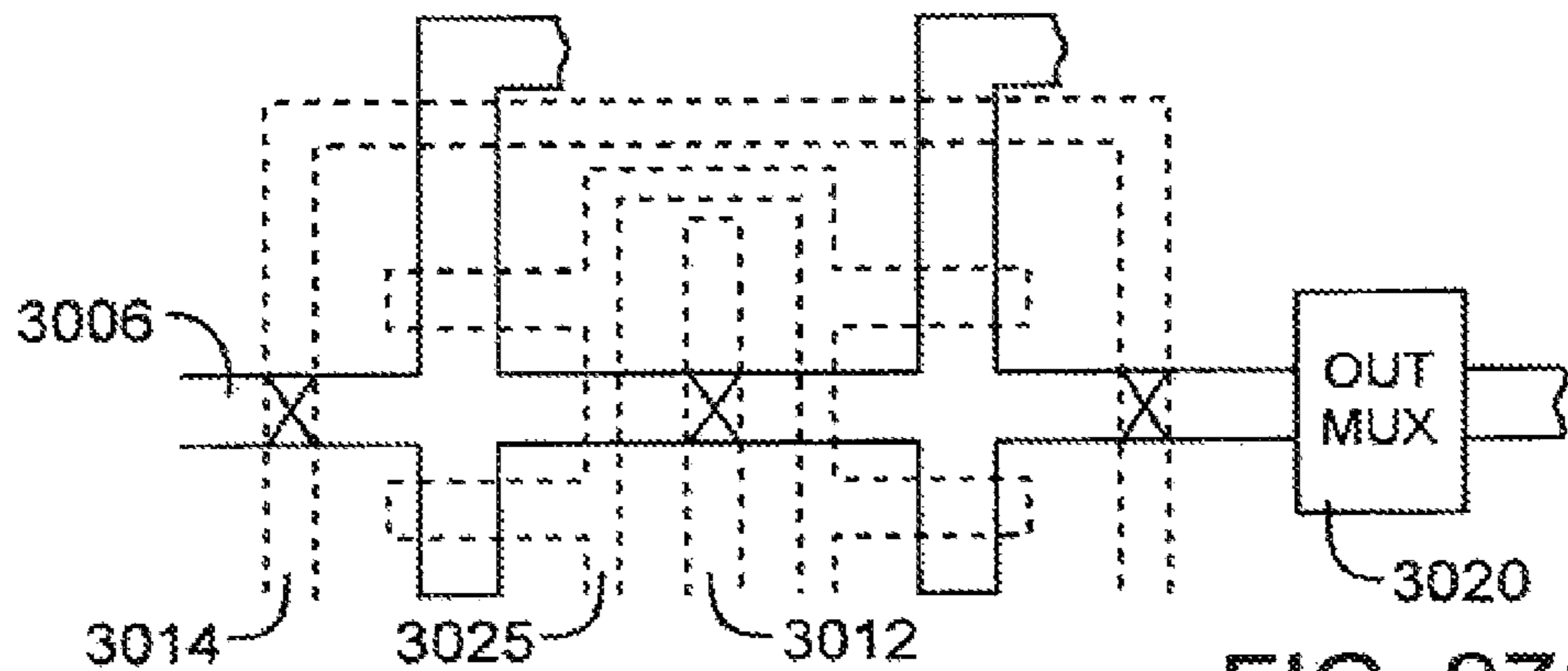


FIG. 27C

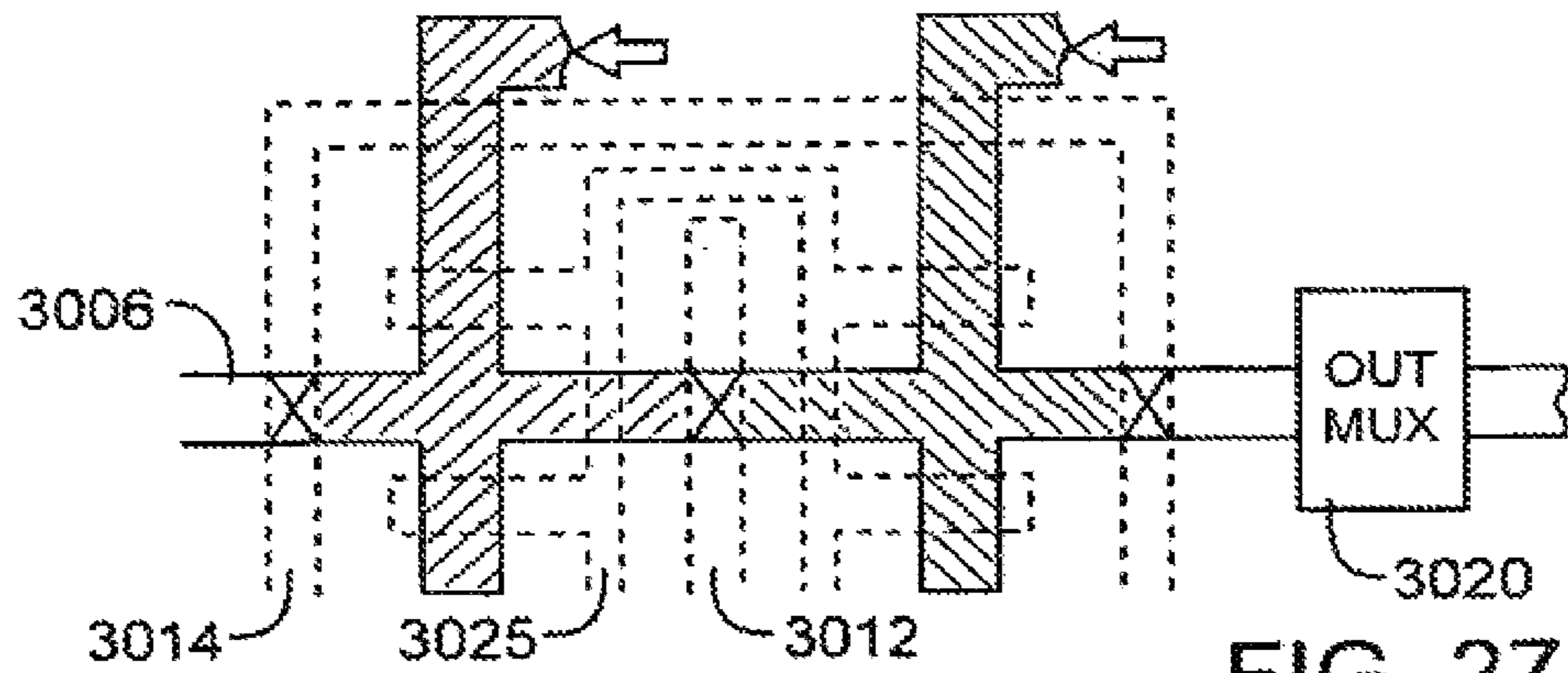


FIG. 27D

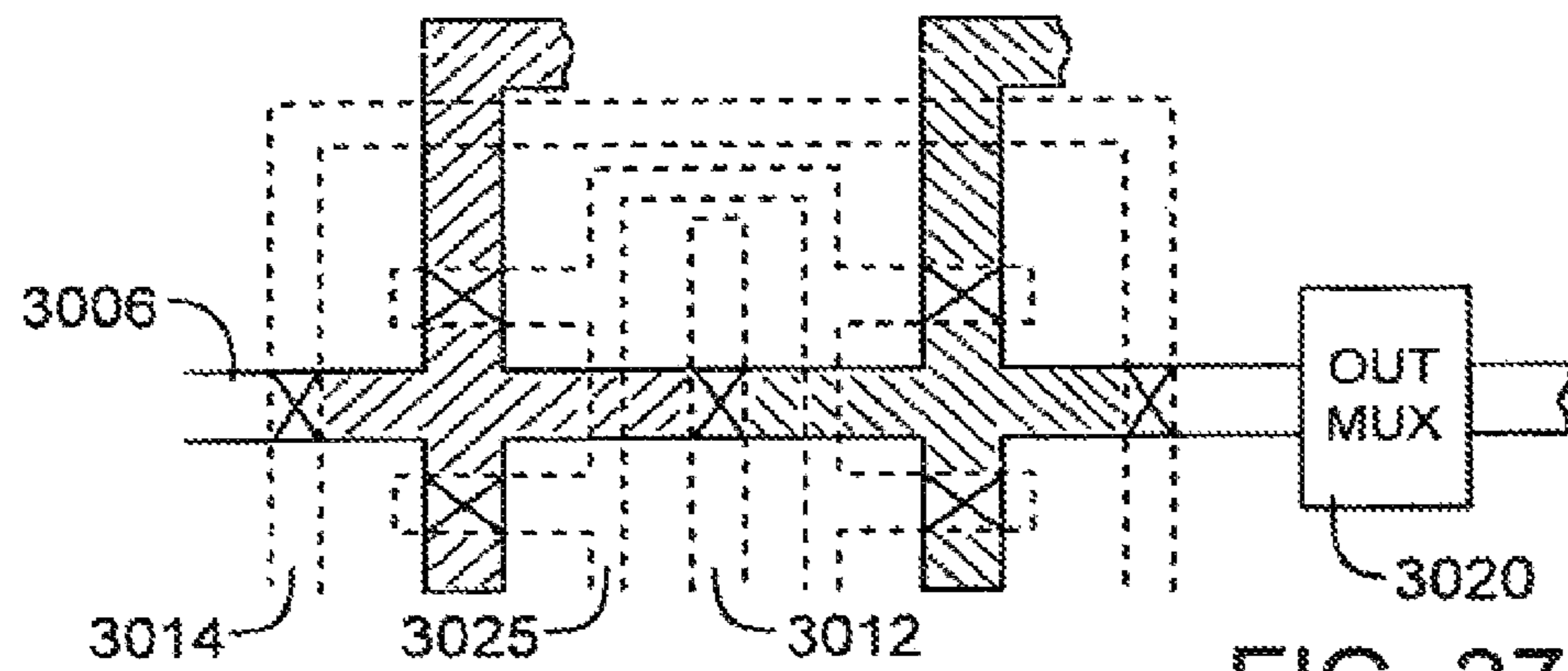


FIG. 27E

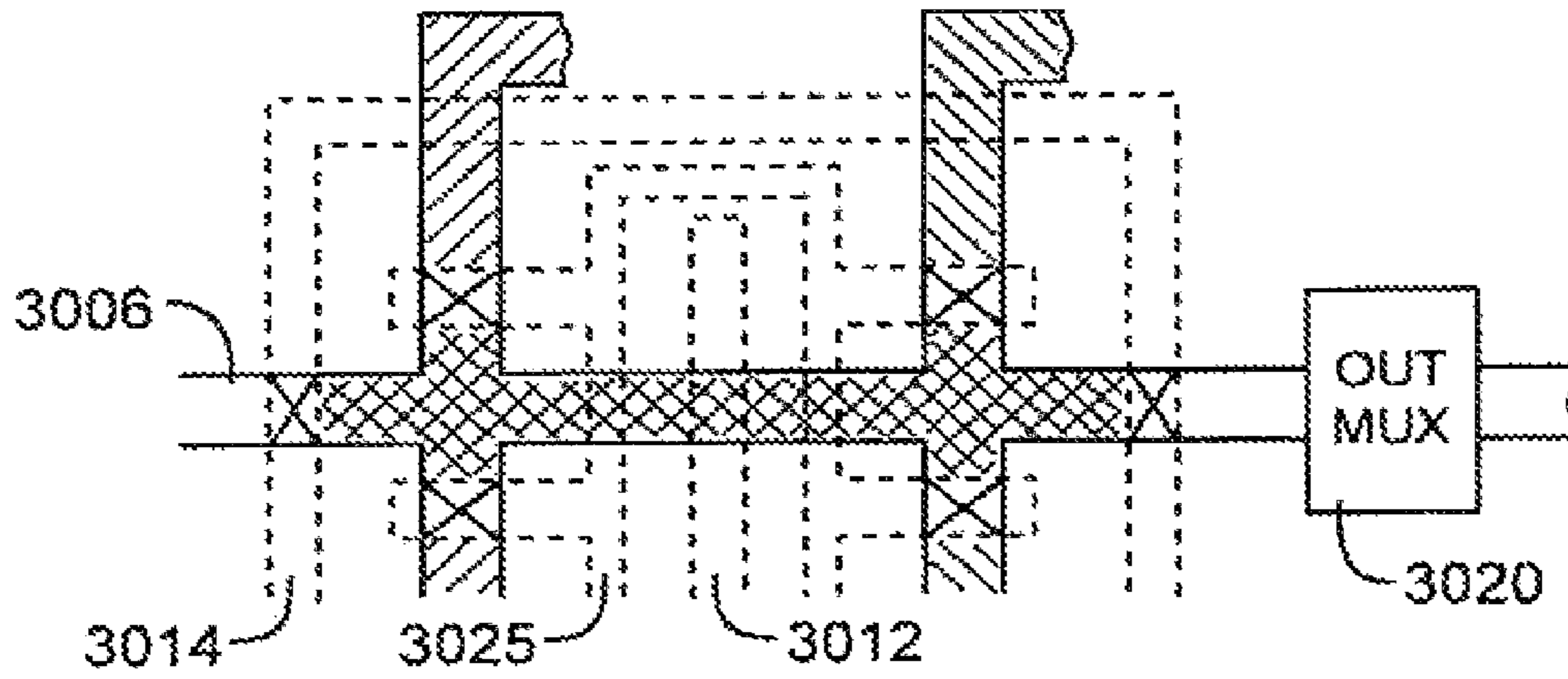


FIG. 27F

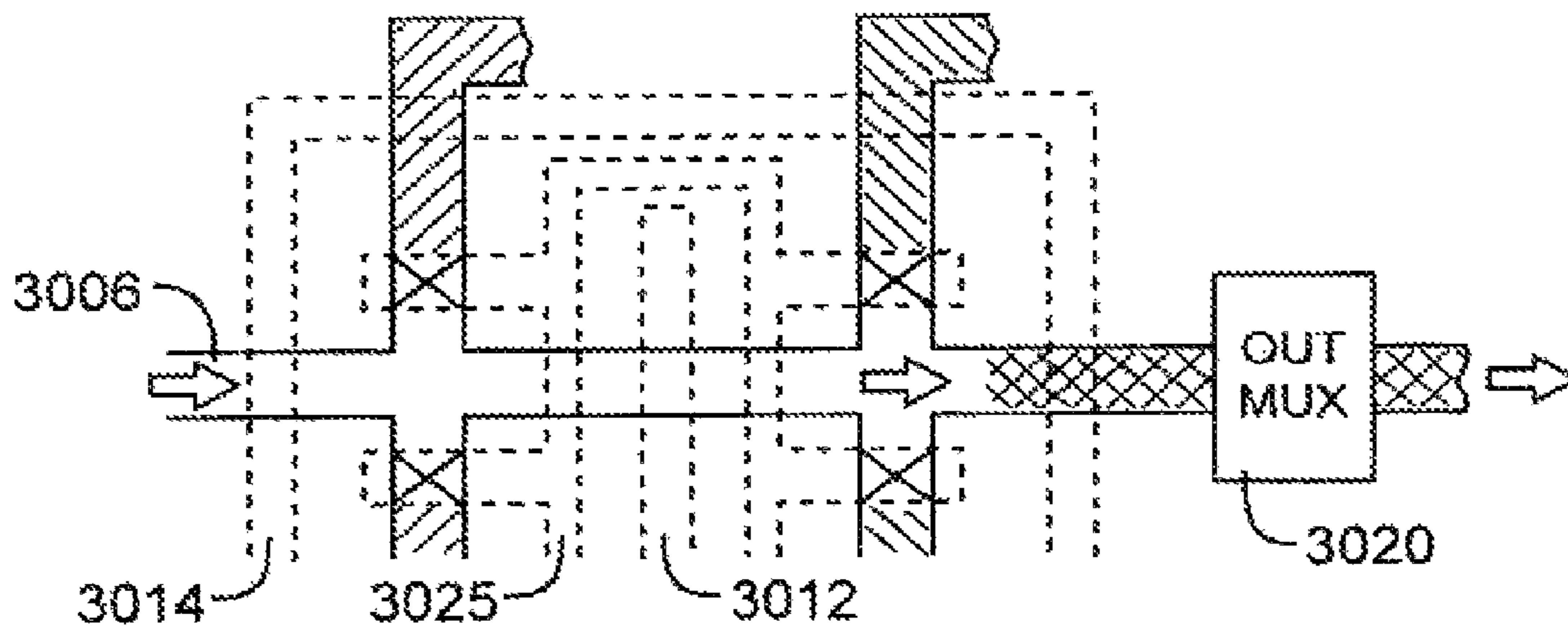
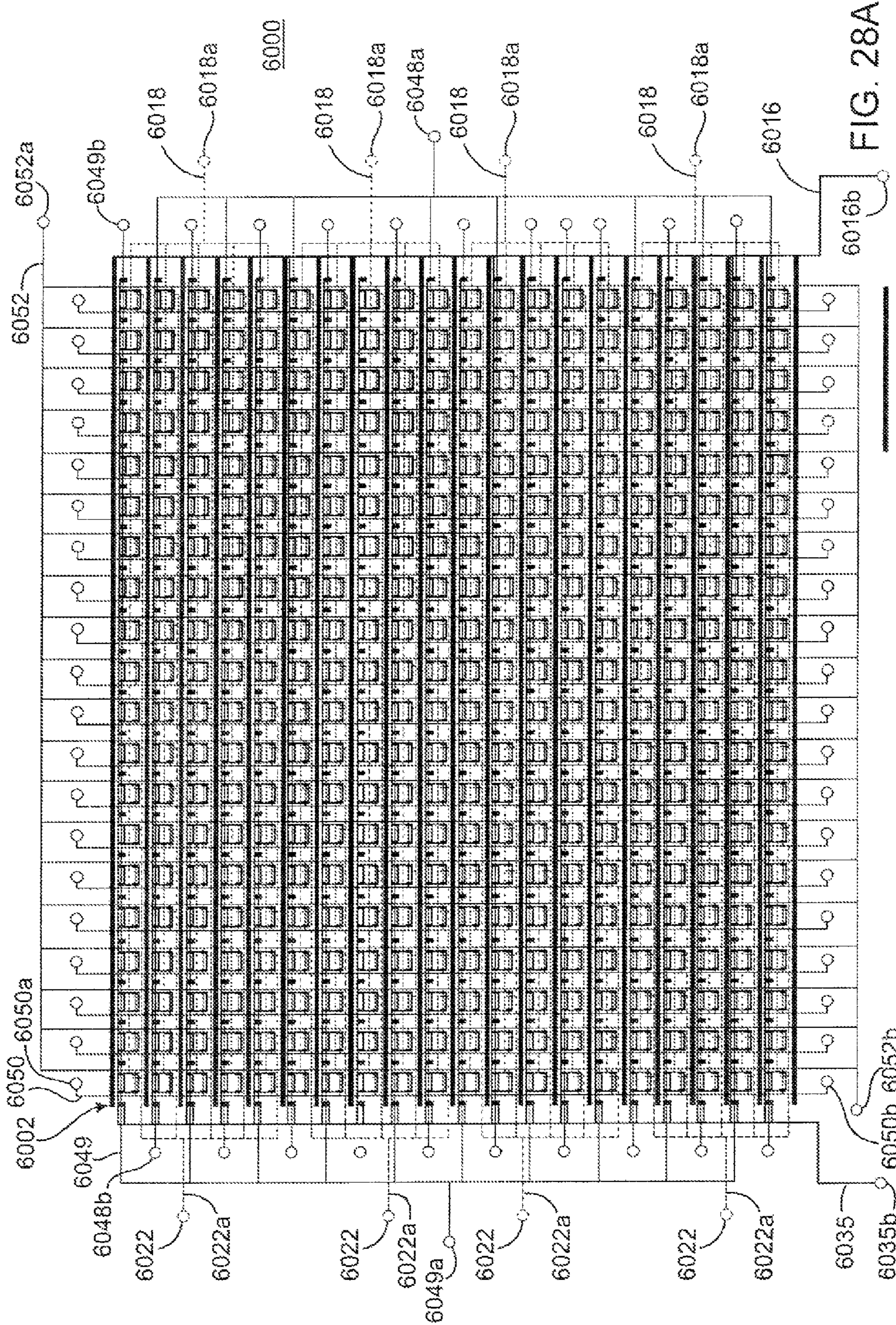


FIG. 27G



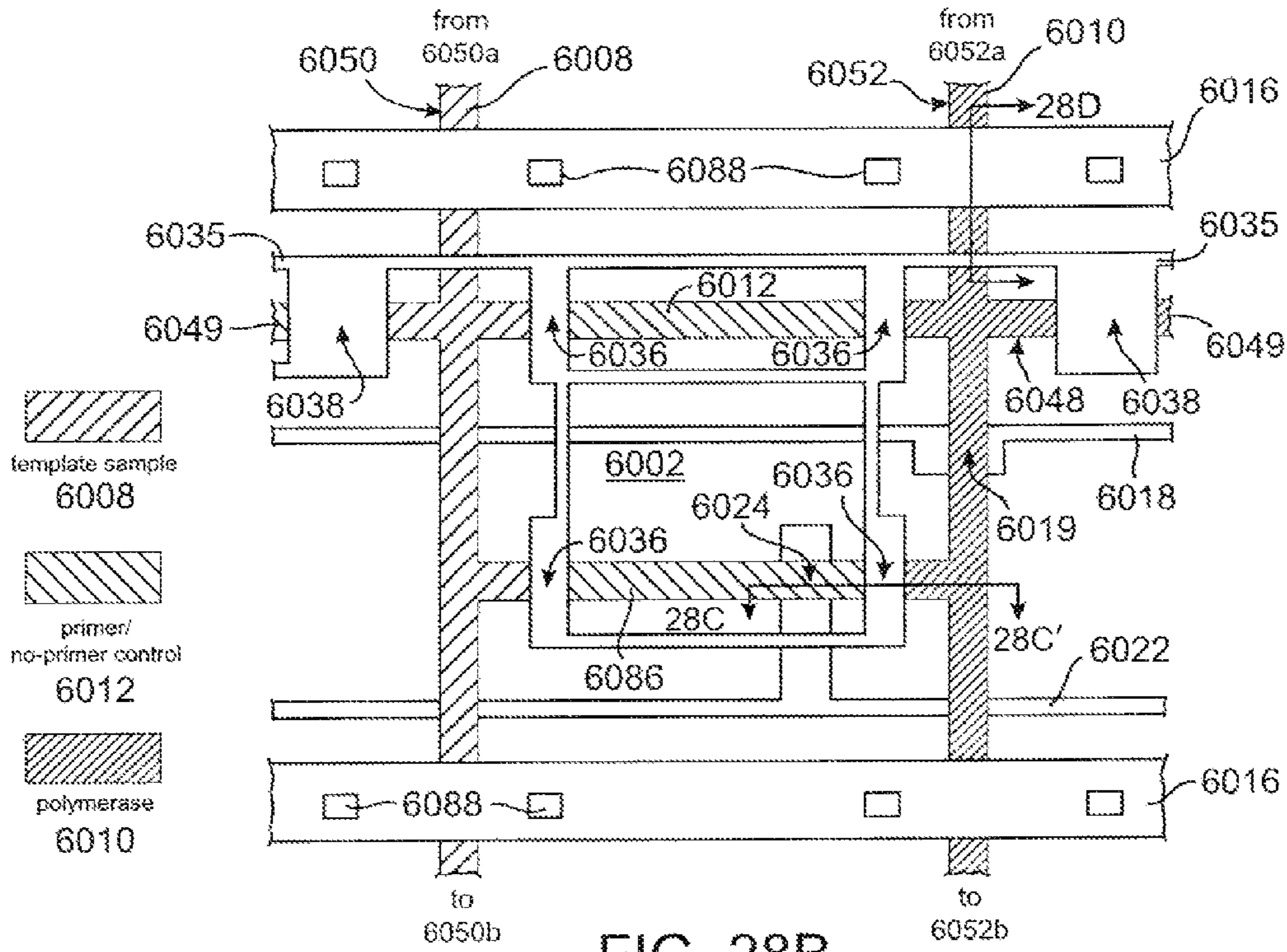


FIG. 28B

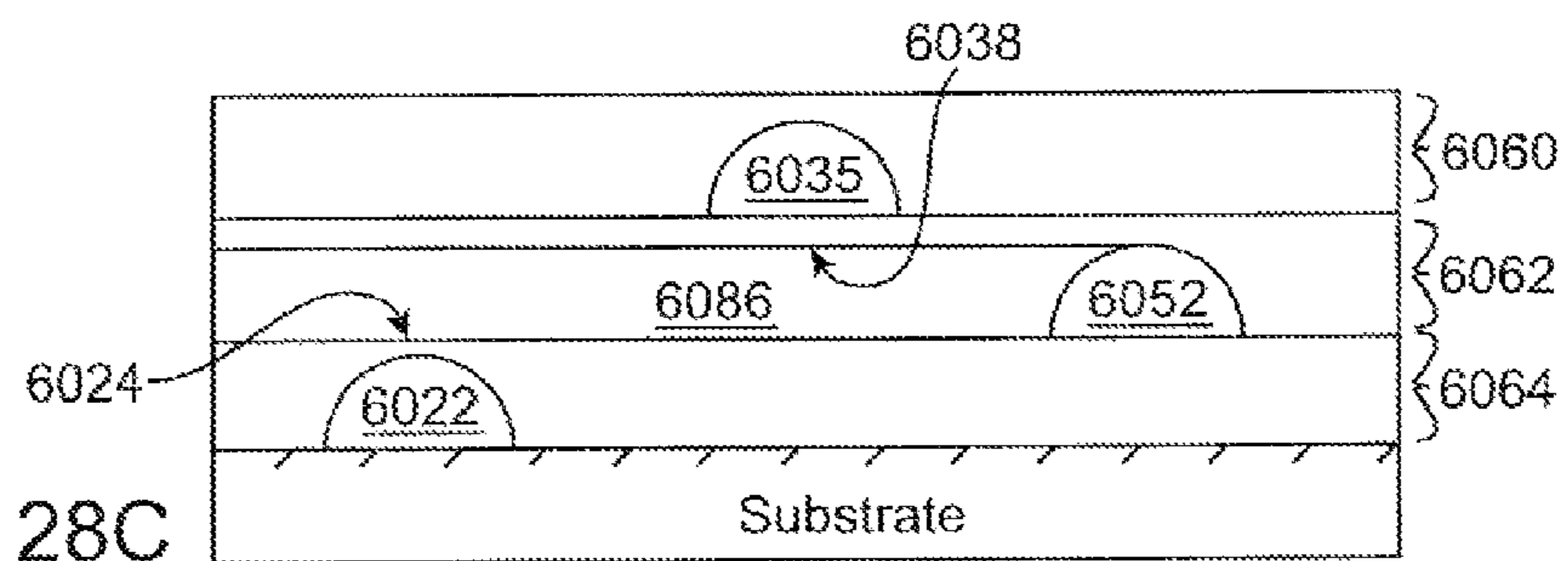


FIG. 28C

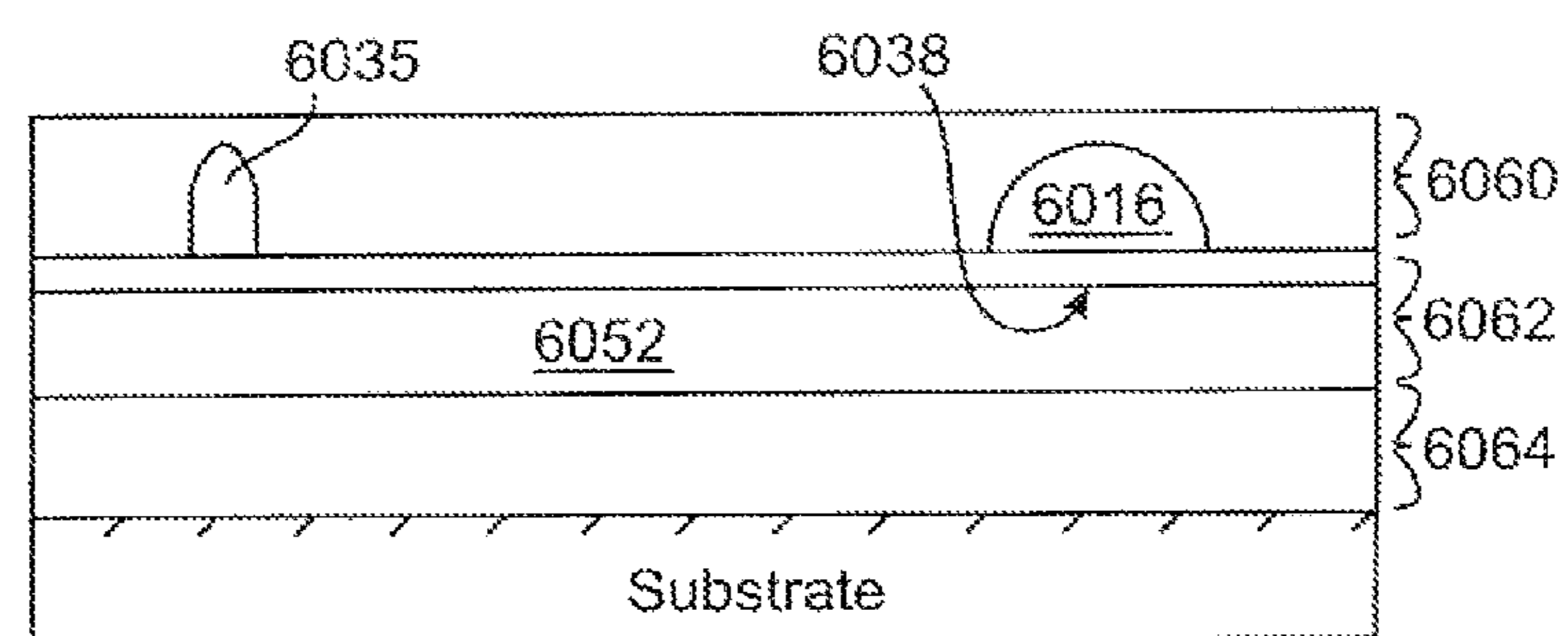


FIG. 28D

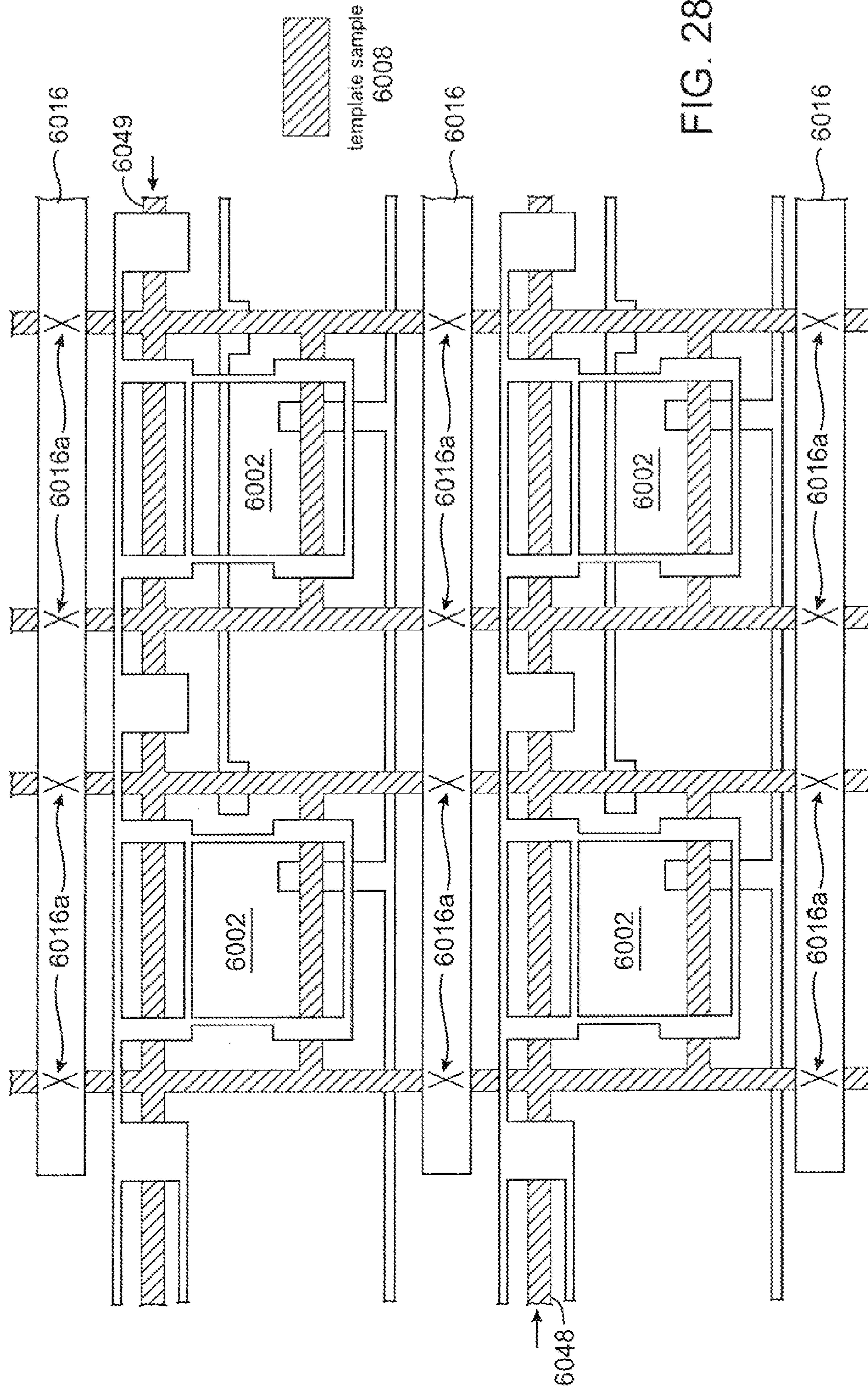


FIG. 28E

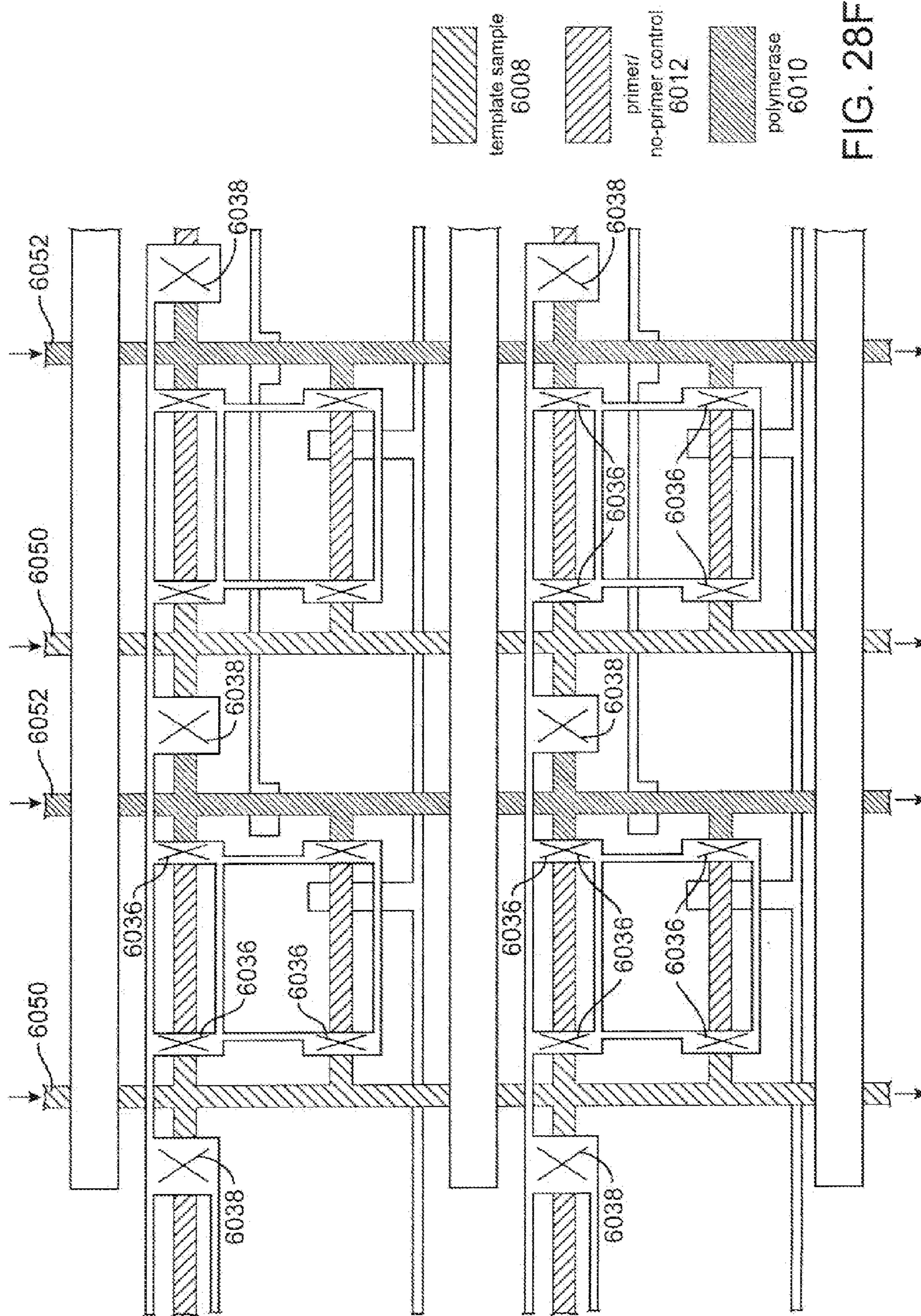


FIG. 28F

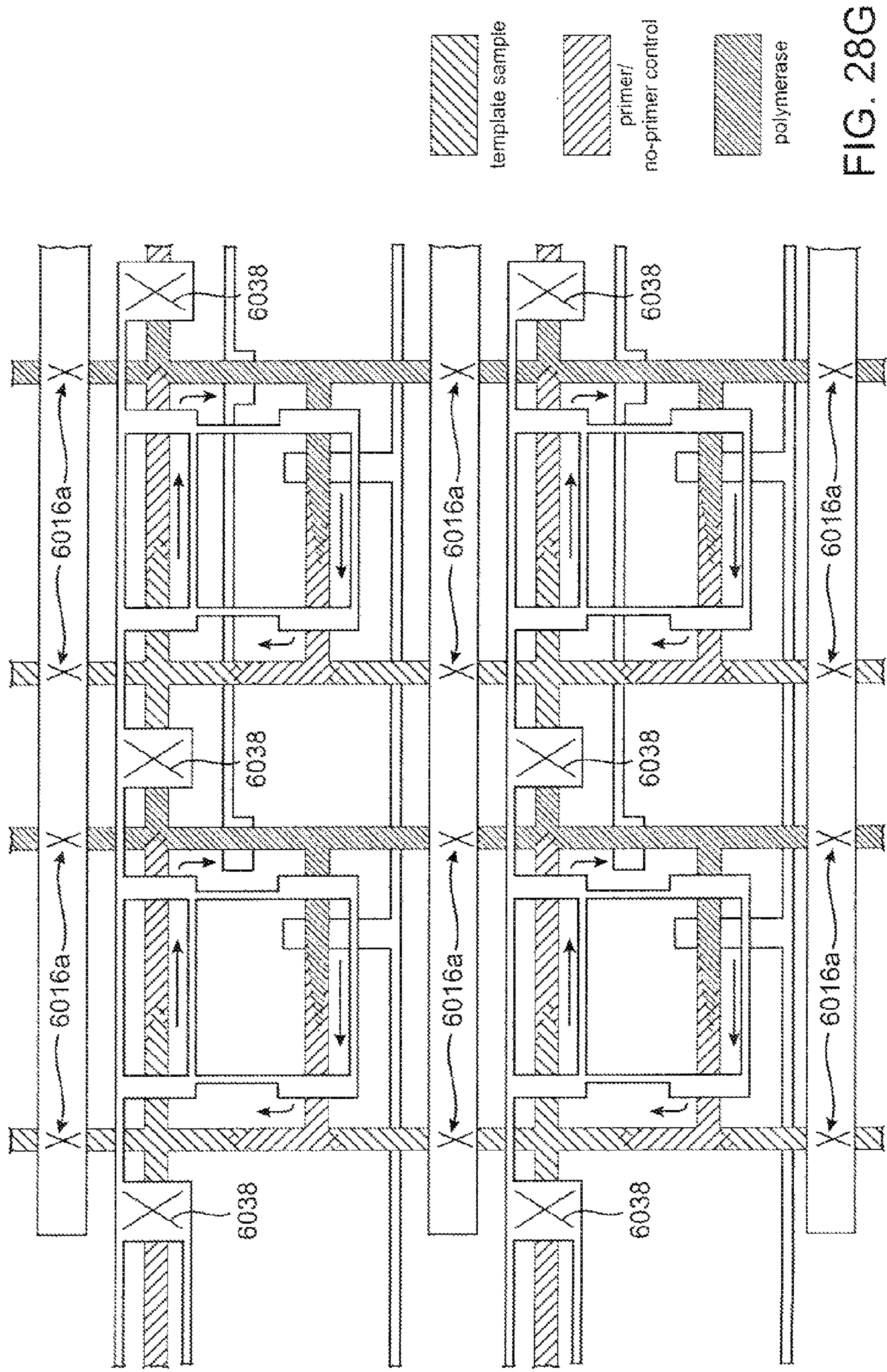


FIG. 28G

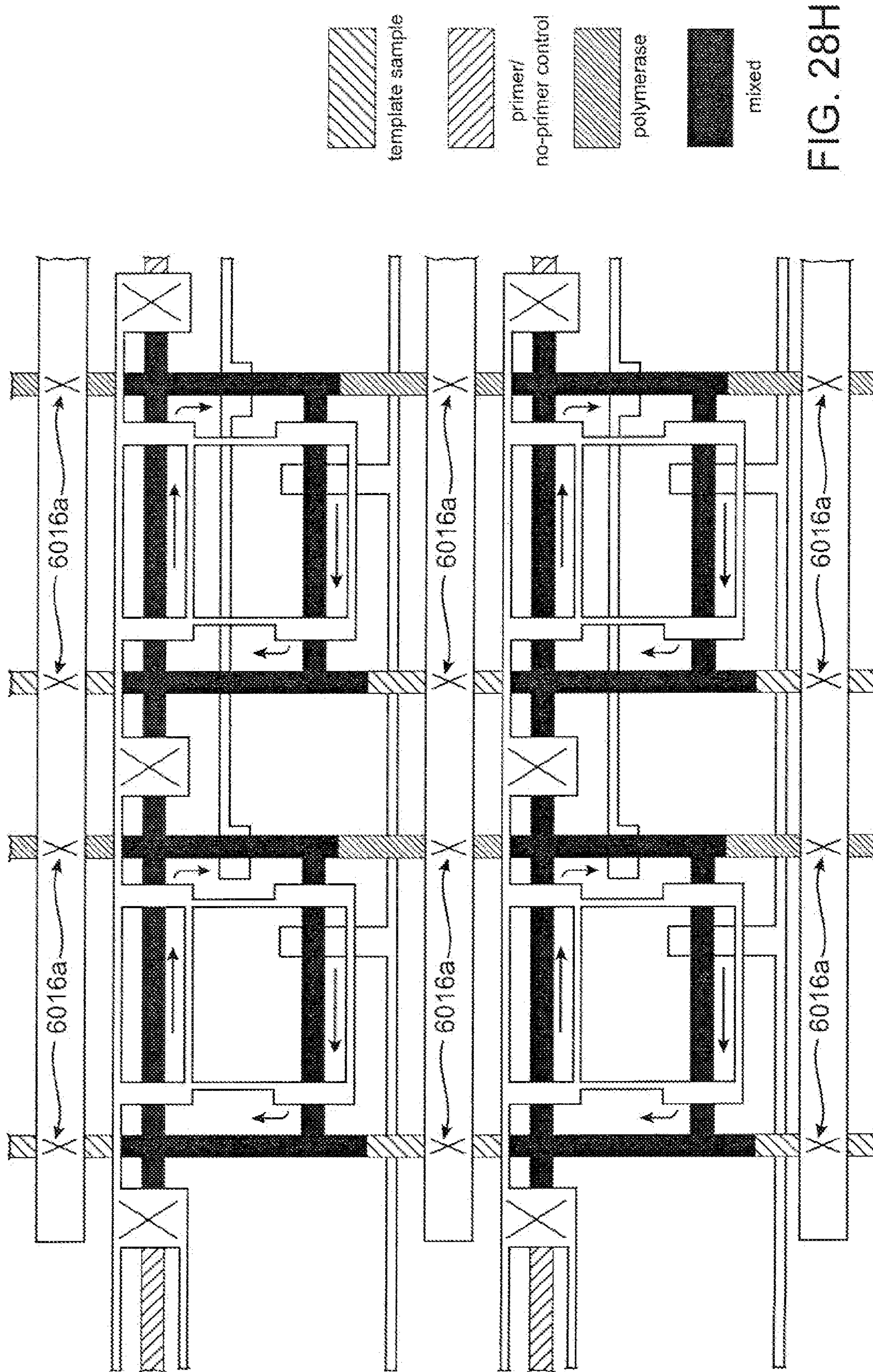


FIG. 28H

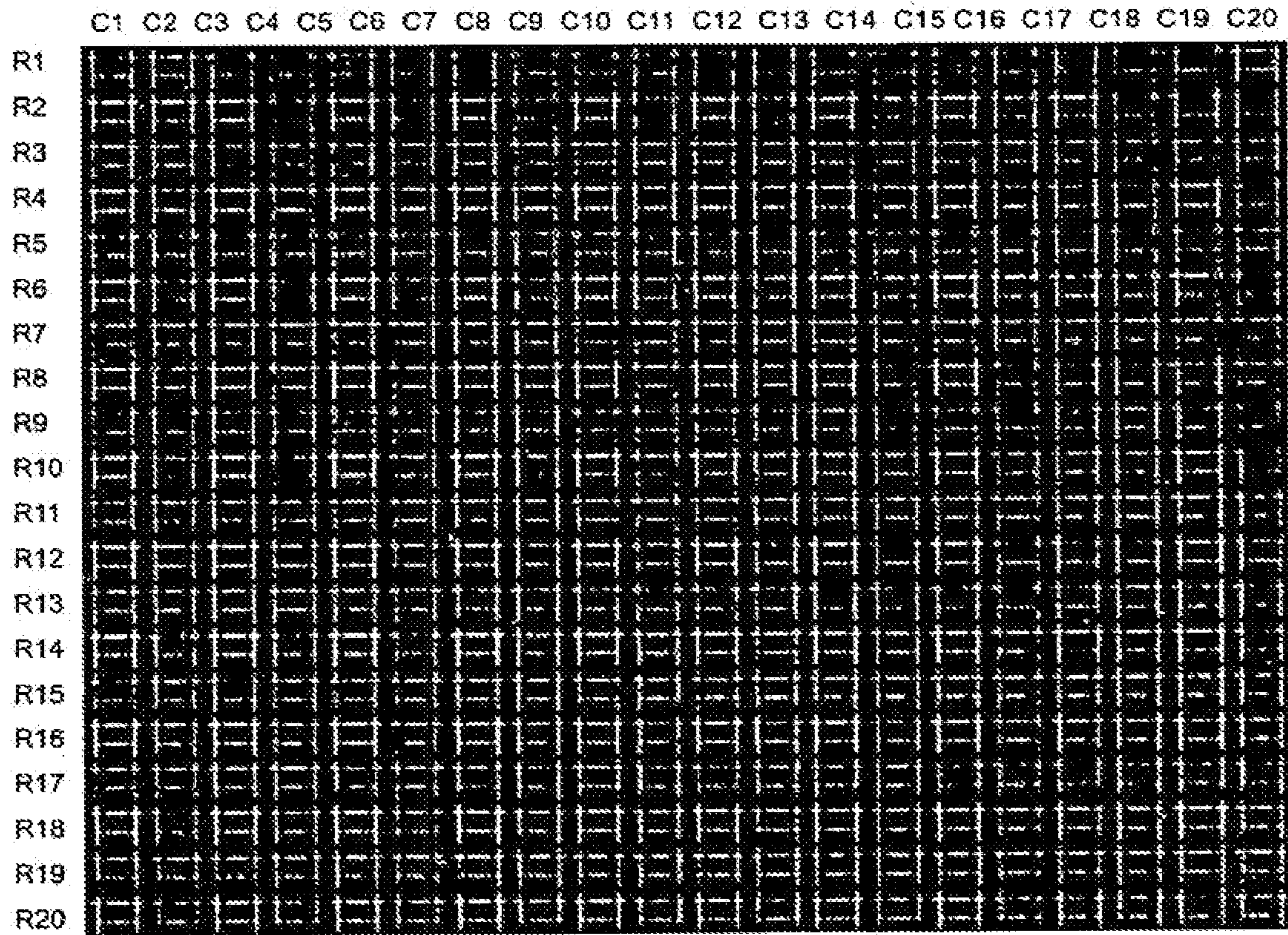


FIG. 29

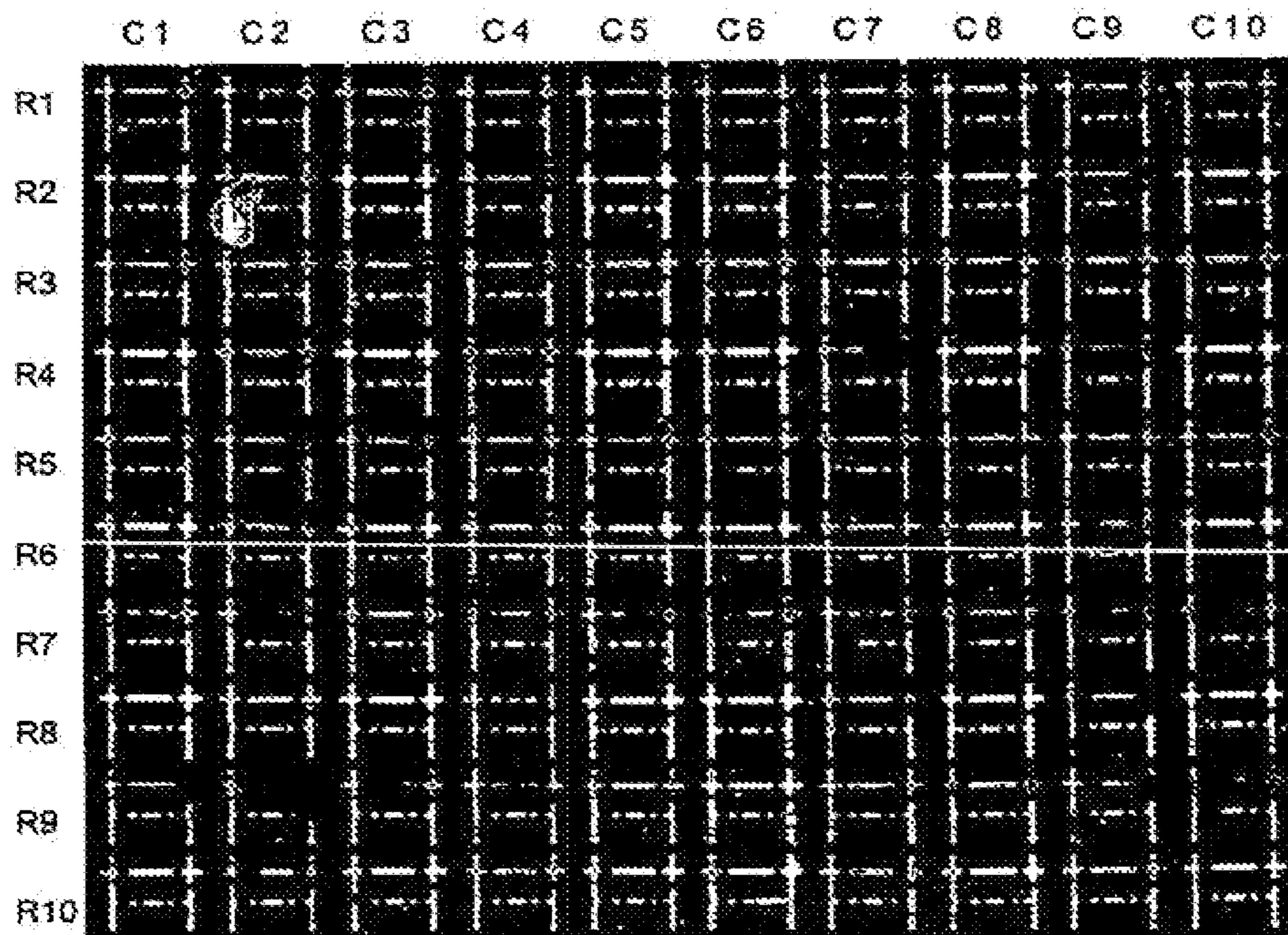


FIG. 30

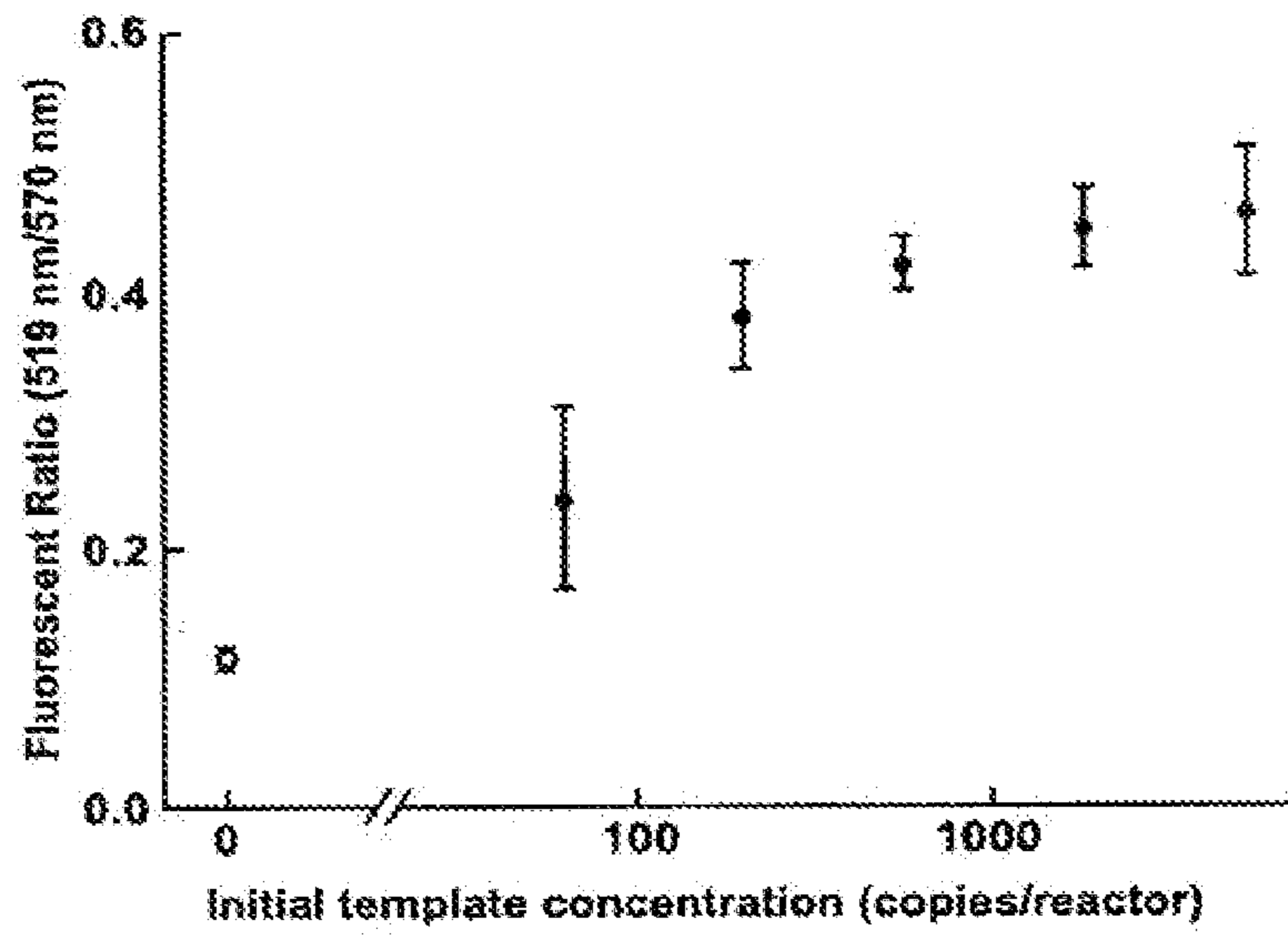


FIG. 31

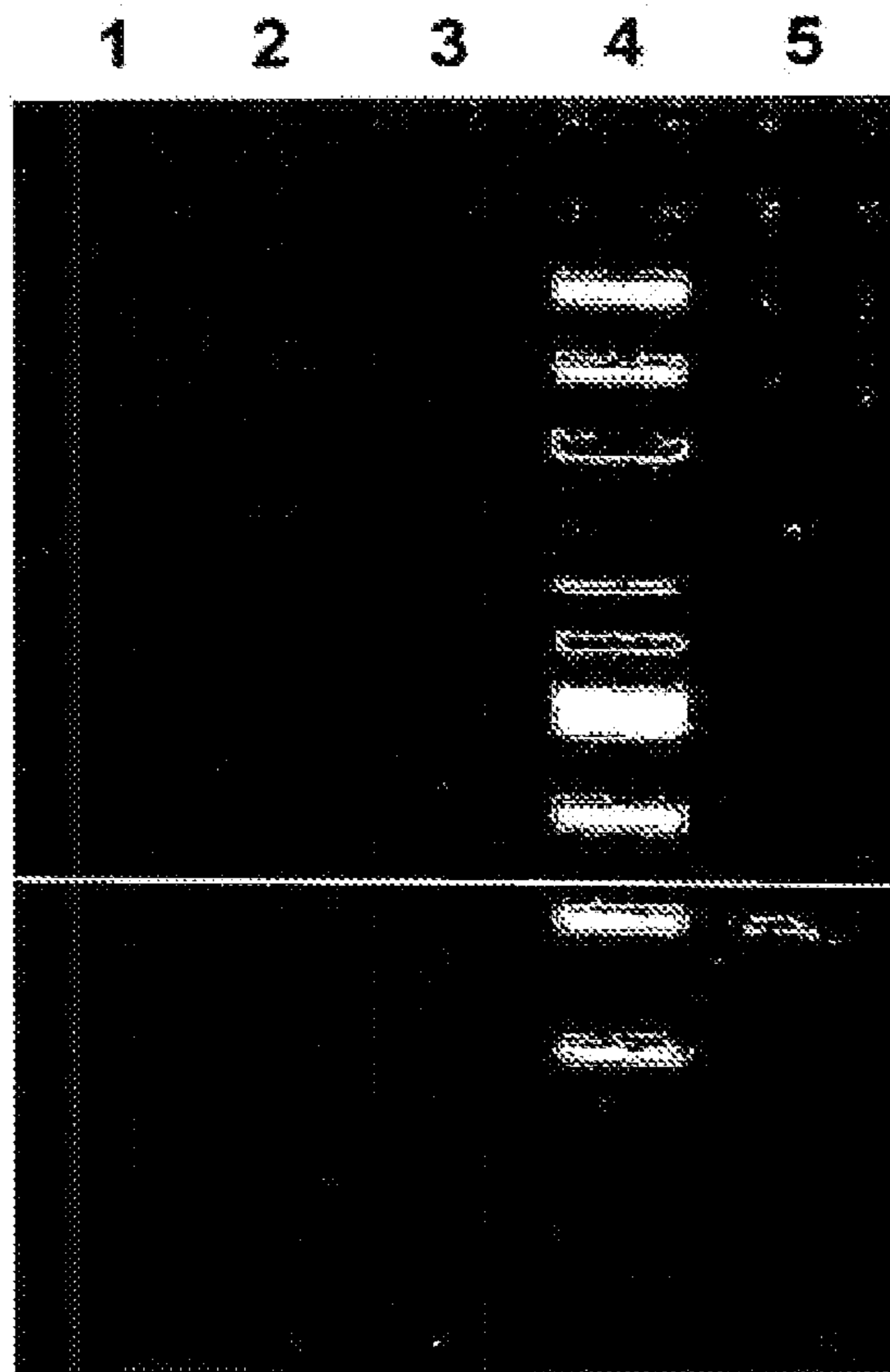


FIG. 32

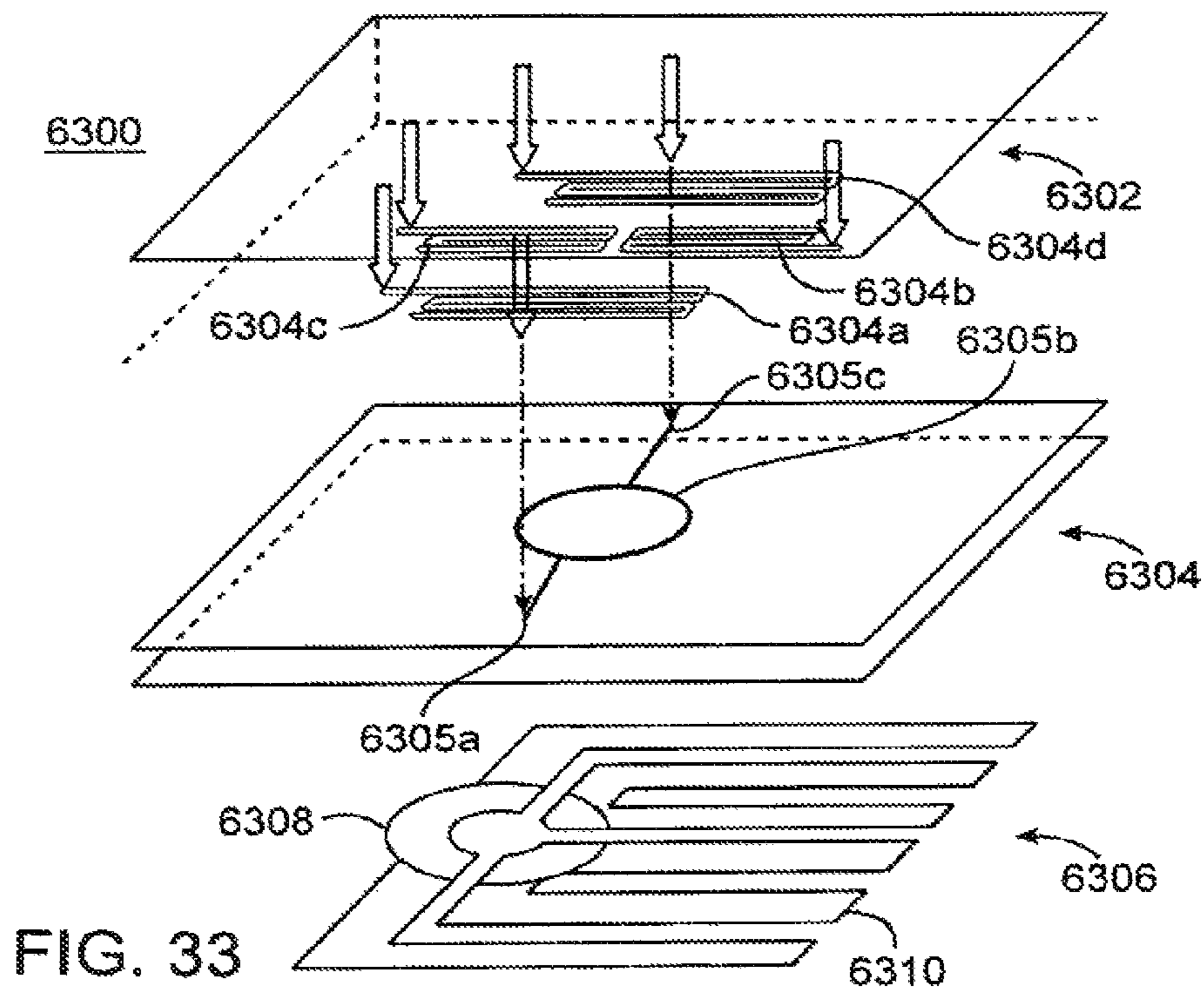


FIG. 33

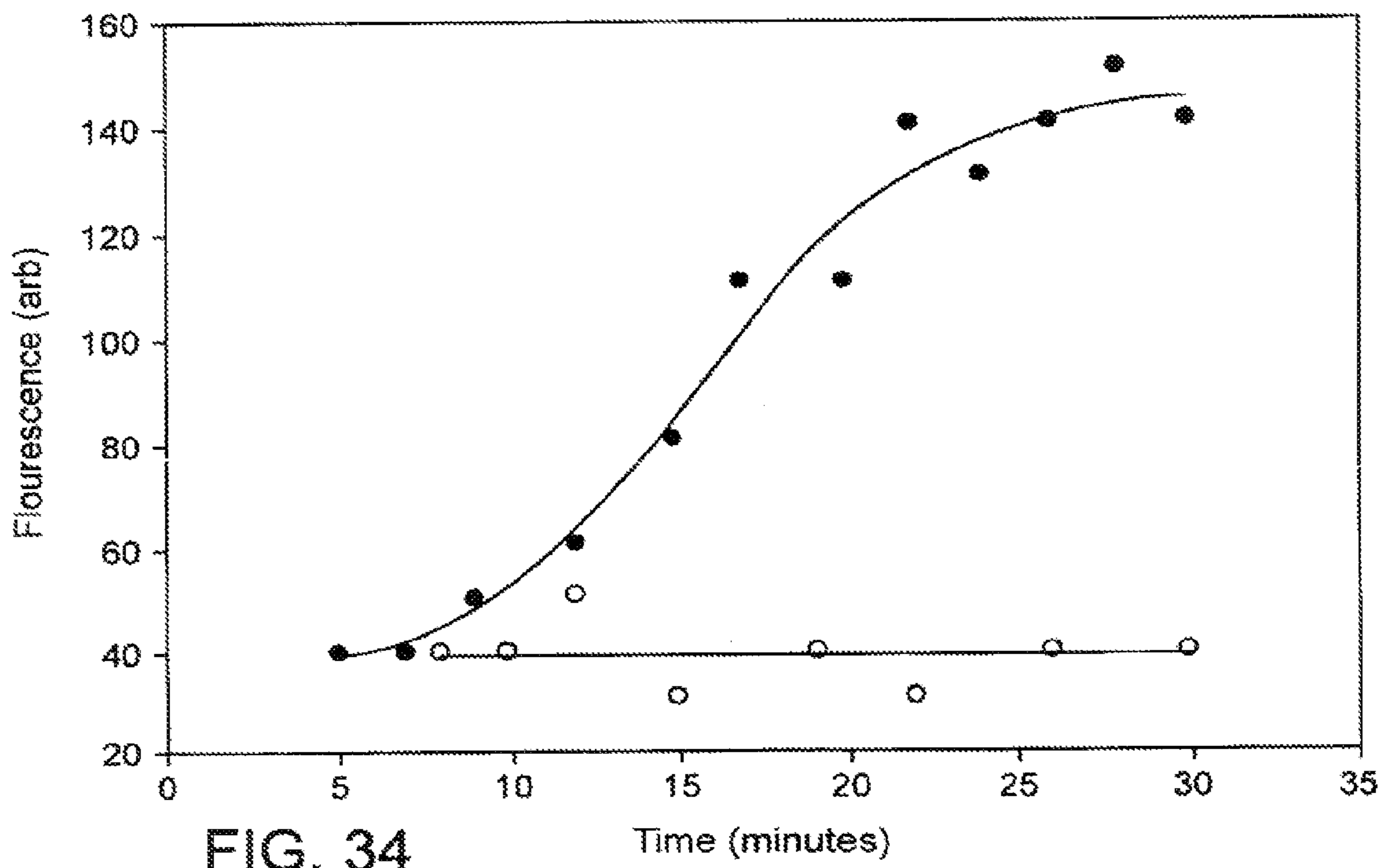
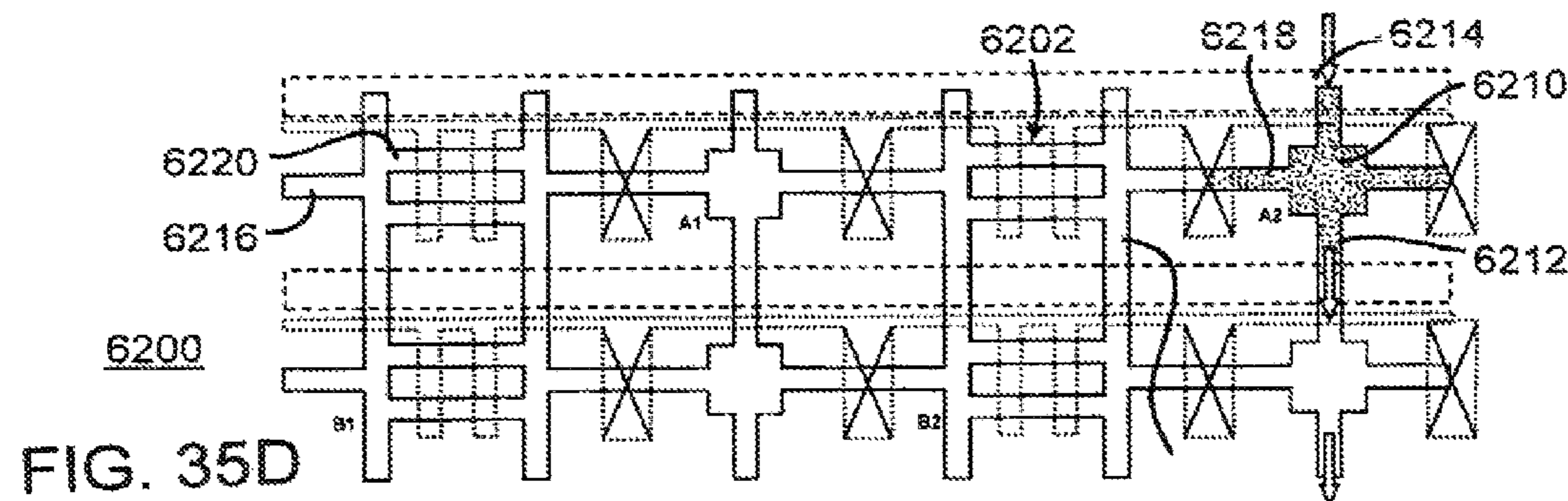
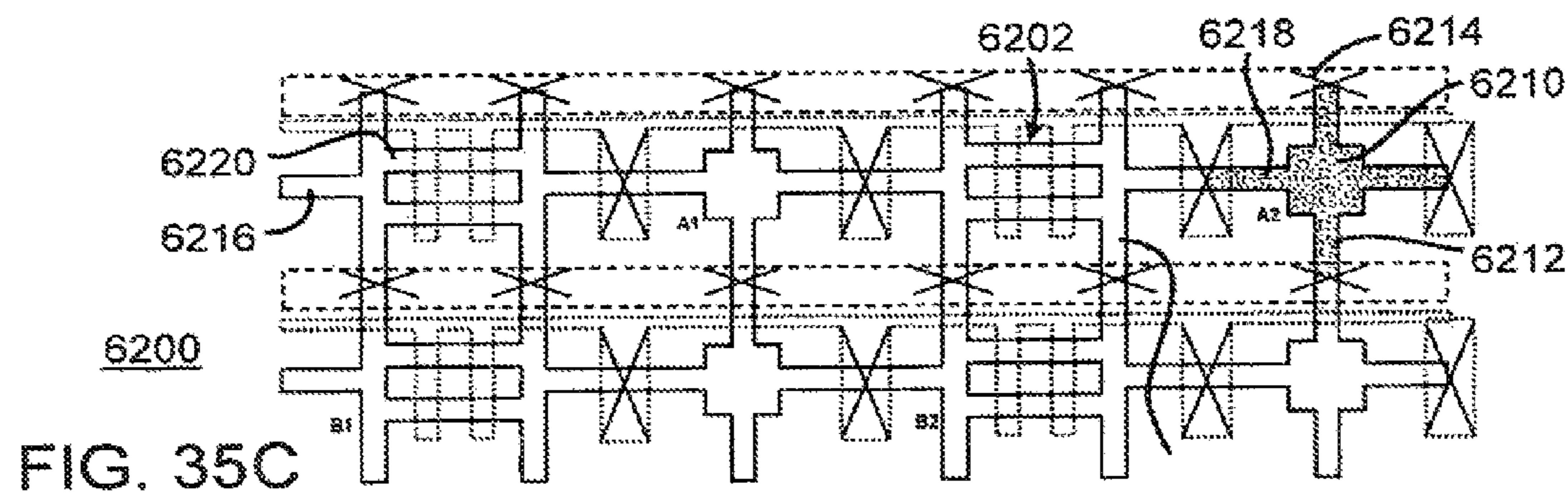
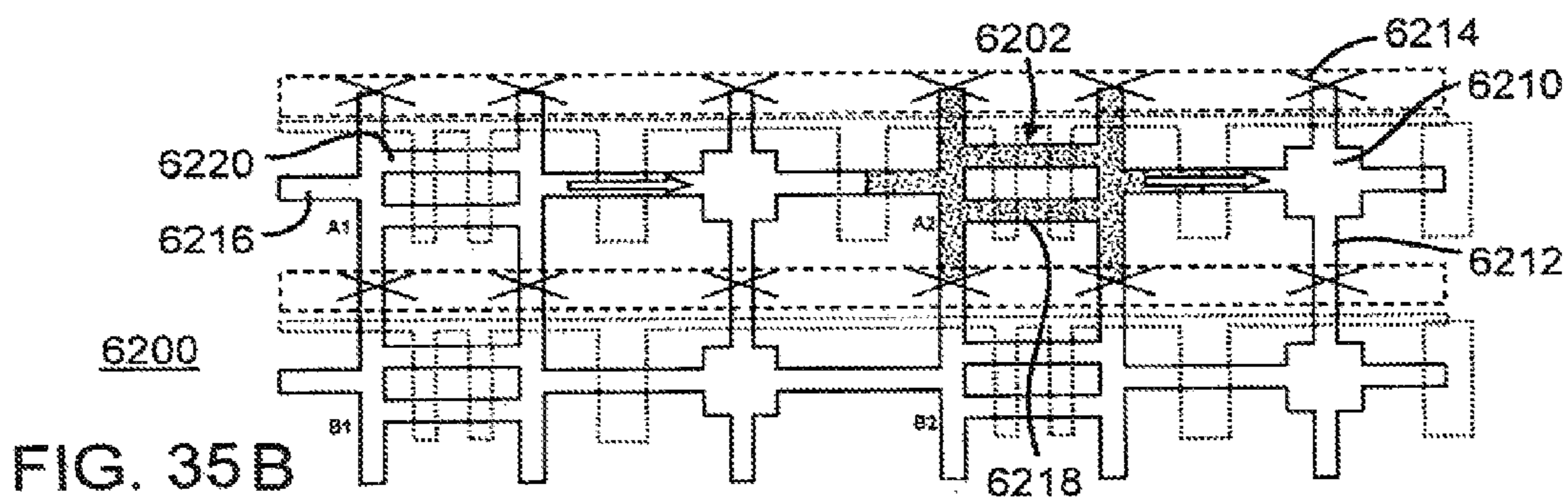
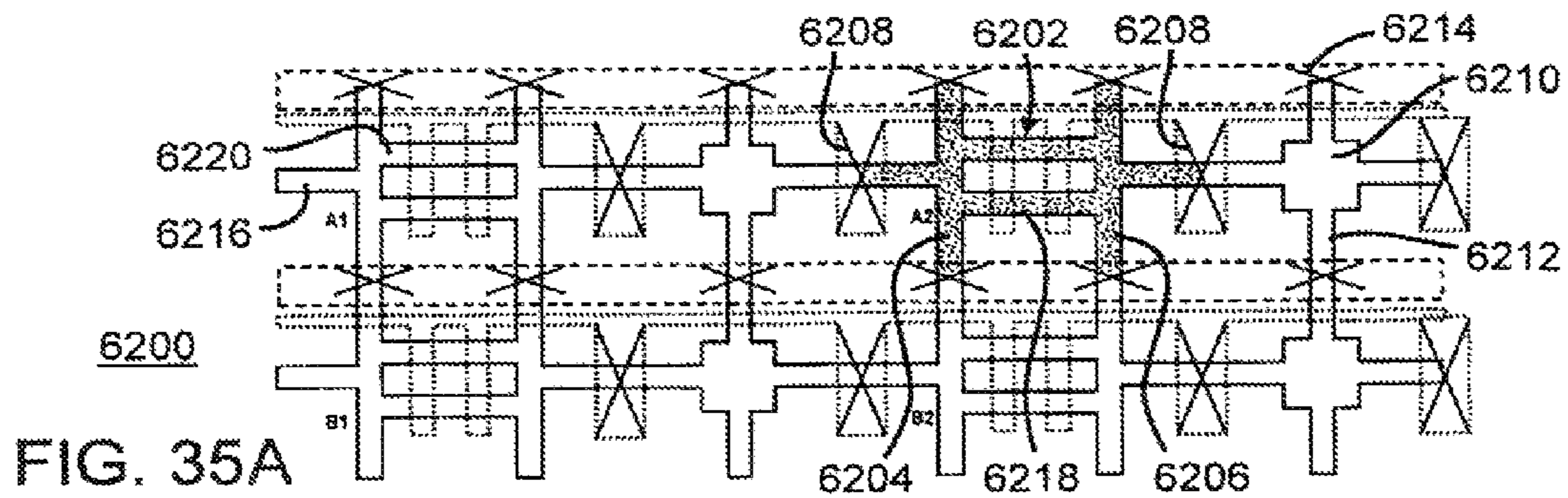


FIG. 34



MICROFLUIDIC ROTARY FLOW REACTOR MATRIX

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is a continuation of U.S. patent application Ser. No. 10/837,025 filed Apr. 30, 2004, entitled "MICROFLUIDIC ROTARY FLOW REACTOR MATRIX," which claims priority to U.S. Provisional Patent Application No. 60/494,432 filed Aug. 11, 2003, the entire disclosures of which are hereby incorporated by reference, for all purposes, as if fully set forth herein.

STATEMENT AS TO RIGHTS TO INVENTIONS MADE UNDER FEDERALLY SPONSORED RESEARCH AND DEVELOPMENT

Work described herein has been supported, in part, by the NSF XYZ on a chip program and by the DARPA Bioflips program. The United States Government may therefore have certain rights in the invention.

BACKGROUND OF THE INVENTION

As described by at least Mitchell in *Nat. Biotechnol.* 19, 717-721 (2001), incorporated by reference herein for all purposes, microfluidic technology offers many possible benefits in chemistry, biology and medicine. One important possible benefit is to automate rote work while reducing the consumption of expensive reagents to the nanoliter or sub-nanoliter scale.

As described in *Anal. Chem.* 69, 3407-3412 (1997), incorporated by reference herein for all purposes, Hadd et al. developed a microchip for performing automated enzyme assay, in which precise concentrations of substrate, enzyme, and inhibitor were mixed in nanoliter volumes using electrokinetic flow. In *Anal. Chem.* 74, 2451-2457 (2002), incorporated by reference herein for all purposes, Fu et al reported an integrated cell sorter with a 1 picoliter minimum active volume by the actuated valve.

However, as pointed out by Meldrum et al. in *Science* 297, 1197-1198 (2002), incorporated by reference herein for all purposes, since microfluidic devices must at some point be interfaced to the macroscopic world, there is a minimum practical volume (of order—1 μ l) that can be introduced into a device. The so-called "world-to-chip" interface problem described by Ramsey in *Nat. Biotechnol.* 17, 1061-1062 (1999), incorporated by reference herein for all purposes, has plagued the microfluidic field since its inception. As noted by Ross et al. in *Anal. Chem.* 74, 2556-2564 (2002), it is always questionable how the desired economies of scale in microfluidics can practically be achieved, unless an effective approach is developed to solve the mismatch between those two scales.

Although integrated glass capillaries have been used to reduce sample consumption for simple titration between two reagents, these devices are fundamentally serial and have the possibility of sample cross-contamination during the loading process. See Farinas et al., *Analytical Biochemistry* 295, 138-142 (2001), incorporated by reference herein for all purposes.

On the other hand, there has been an effort to develop techniques to concentrate analytes from a large input volume. In *Anal. Chem.* 73, 1627-1633 (2001), incorporated by reference herein for all purposes, Macounova et al. describe microfluidic isoelectric focusing (IEF) techniques. Ross et al.

describe temperature gradient focusing (TGF) techniques. However, only a limited species of samples have successfully been demonstrated so far.

The challenge associated with realizing the desired economies of scale in microfluidic devices is to simultaneously reduce the number of pipetting steps needed to load the devices, while amortizing the sample volume from each pipetting step over a large number of independent assays. As pointed out in U.S. Pat. No. 6,508,988, incorporated by reference herein for all purposes, microfluidic matrix geometries offer the advantage of performing N^2 independent reactions with only $2N$ pipetting steps.

In *Lab on a Chip* 2, 188-192 (2002), Kikutani et al. used $N=2$ matrices for chemical synthesis. In *Anal. Chem.*, 73, 5207-5213 (2001), Ismagilov et al. used $N=5$ passive matrices to demonstrate two-component biochemical assays such as optical detection of enzymatic activity. However, in those devices the reagent consumption scaled only with N , which was so small that there were little practical savings.

Passive devices also have technical limitations in sample metering and device operation. For example, precise pressure balancing was required during operation of the devices of Kikutani et al., and the kinetics of mixing were limited due to the static nature in the devices of Ismagilov et al.

Accordingly, there is a need in the art for microfluidic techniques and apparatuses for addressing the "world-to-chip" interface problem.

SUMMARY OF THE INVENTION

An embodiment of a microfluidic matrix device in accordance with the present invention offers an effective solution to the "world-to-chip" interface problem by accomplishing two important goals simultaneously: an economy of scale in reagent consumption is achieved, while simultaneously minimizing pipetting steps. N^2 independent assays can be performed with only $2N+1$ pipetting steps, using a single aliquot of the enzyme that can be amortized over all the reactors. Thus, the chip reduces labor relative to conventional fluid handling techniques by using an order of magnitude less pipetting steps and reduces cost by consuming two to three orders of magnitude less reagents per reaction. The demonstrated PCR format has immediate applications in medical diagnosis and gene testing. Beyond PCR, the microfluidic matrix chip provides a universal and flexible platform for biological and chemical assays that require parsimonious use of precious reagents and highly automated processing.

An embodiment of a microfluidic device in accordance with the present invention, comprises, a plurality of flow channels defined within an elastomer layer to form a matrix of rotary flow reactors. A first set of control lines are proximate to and separated from the flow channels by first elastomer membranes, the first elastomer membranes actuatable to introduce fluids into the rotary flow reactors and to isolate the rotary flow reactors. A second set of control lines are proximate to and separated from the flow channels by second elastomer membranes, the second elastomer membranes actuatable to cause peristaltic pumping of the fluids within the isolated rotary flow reactors.

A method of conducting a chemical reaction, the method comprising, providing a microfluidic device comprising a plurality of flow channels defined within an elastomer layer to form a matrix of rotary flow reactors, a first set of control lines proximate to and separated from the flow channels by first elastomer membranes, and a second set of control lines proximate to and separated from the flow channels by second elastomer membranes. First and second chemicals are intro-

duced to the rotary flow reactors. Pressure is applied to the first set of control lines to actuate the first elastomer membranes to isolate the rotary flow reactors, and pressure is applied to the second set of control lines to actuate the second elastomer membranes to cause peristaltic pumping of the fluids within the isolated rotary flow reactors.

These and other embodiments of the present invention, as well as its advantages and features, are described in more detail in conjunction with the text below and attached figures.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is an illustration of a first elastomeric layer formed on top of a micromachined mold.

FIG. 2 is an illustration of a second elastomeric layer formed on top of a micromachined mold.

FIG. 3 is an illustration of the elastomeric layer of FIG. 2 removed from the micromachined mold and positioned over the top of the elastomeric layer of FIG. 1.

FIG. 4 is an illustration corresponding to FIG. 3, but showing the second elastomeric layer positioned on top of the first elastomeric layer.

FIG. 5 is an illustration corresponding to FIG. 4, but showing the first and second elastomeric layers bonded together.

FIG. 6 is an illustration corresponding to FIG. 5, but showing the first micromachined mold removed and a planar substrate positioned in its place.

FIG. 7A is an illustration corresponding to FIG. 6, but showing the elastomeric structure sealed onto the planar substrate.

FIG. 7B is a front sectional view corresponding to FIG. 7A, showing an open flow channel.

FIGS. 7C-7G are illustrations showing steps of a method for forming an elastomeric structure having a membrane formed from a separate elastomeric layer.

FIG. 7H is a front sectional view showing the valve of FIG. 7B in an actuated state.

FIGS. 8A and 8B illustrates valve opening vs. applied pressure for various flow channels.

FIG. 9 illustrates time response of a $100\ \mu\text{m}\times 100\ \mu\text{m}\times 10\ \mu\text{m}$ RTV microvalve.

FIG. 10 is a front sectional view of the valve of FIG. 7B showing actuation of the membrane.

FIG. 11 is a front sectional view of an alternative embodiment of a valve having a flow channel with a curved upper surface.

FIG. 12A is a top schematic view of an on/off valve.

FIG. 12B is a sectional elevation view along line 23B-23B in FIG. 12A.

FIG. 13A is a top schematic view of a peristaltic pumping system.

FIG. 13B is a sectional elevation view along line 24B-24B in FIG. 13A.

FIG. 14 is a graph showing experimentally achieved pumping rates vs. frequency for an embodiment of the peristaltic pumping system of FIG. 13.

FIG. 15A is a top schematic view of one control line actuating multiple flow lines simultaneously.

FIG. 15B is a sectional elevation view along line 26B-26B in FIG. 15A.

FIG. 16 is a schematic illustration of a multiplexed system adapted to permit flow through various channels.

FIGS. 17A-D show plan views of one embodiment of a switchable flow array.

FIGS. 18A-D show plan views of one embodiment of a cell pen array structure.

FIGS. 19A-19B show plan and cross-sectional views illustrating operation of one embodiment of a cell cage structure in accordance with the present invention.

FIGS. 20A-20D show plan views of operation of a structure utilizing cross-channel injection in accordance with the embodiment of the present invention.

FIG. 21 shows a plan view of one embodiment of a rotary mixing structure in accordance with the present invention.

FIG. 22A shows a simplified plan view illustrating a binary tree microfluidic multiplexor operational diagram.

FIG. 22B shows a simplified plan view illustrating a tertiary tree microfluidic multiplexor operational diagram.

FIG. 22C shows a simplified cross-sectional view of the general microfluidic architecture of the devices of FIGS. 22A-B.

FIG. 23 plots the number of control lines versus the number of flow lines being controlled, for multiplexors of various base numbers.

FIG. 24 shows a simplified plan view of an embodiment of a microfluidic structure utilizing control channels to control other control channels.

FIG. 24A shows a simplified cross-sectional view of the structure of FIG. 24 taken along the line A-A'.

FIG. 24B shows a simplified cross-sectional view of the structure of FIG. 24 taken along the line B-B'.

FIG. 25 shows a simplified cross-sectional view of the general microfluidic architecture of the device of FIGS. 24-24B.

FIG. 26 shows a simplified plan view of an alternative embodiment of a microfluidic structure utilizing control channels to control other control channels.

FIG. 26A shows a simplified cross-sectional view of the structure of FIG. 26 taken along the line A-A'.

FIG. 26B shows a simplified cross-sectional view of the structure of FIG. 26 taken along the line B-B'.

FIG. 27A is a simplified schematic view of a microfluidic comparator.

FIGS. 27B-G are enlarged simplified plan views showing loading of the chamber of a microfluidic structure of FIG. 27B.

FIG. 28A shows a simplified schematic plan view of one embodiment of a microfluidic matrix architecture in accordance with the present invention.

FIG. 28B shows an enlarged plan view of a single reactor x. FIG. 28C shows a simplified cross-sectional view of the microfluidic matrix of FIG. 28B taken along line 28C-28C'.

FIG. 28D shows a simplified cross-sectional view of the microfluidic matrix of FIG. 28D taken along line 28D-28D'.

FIGS. 28E-H show enlarged and simplified plan views illustrating successive steps of operation of the microfluidic matrix of FIGS. 28A-D.

FIG. 29 shows a two-color image of fluorescent emission from a 20×20 matrix chip.

FIG. 30 shows a scanned fluorescent image illustrating use of the matrix chip to explore combinations of forward and reverse primers.

FIG. 31 plots fluorescent ratio vs. the initial concentration of the cDNA templates.

FIG. 32 shows Agarose gel electrophoresis of the PCR amplicon by the chip or by a microtube.

FIG. 33 shows an exploded view of an alternative embodiment of a microfluidic structure for performing PCR.

FIG. 34 plots fluorescence versus time for nucleic acids amplified by PCR utilizing the microfluidic structure shown in FIG. 34.

FIGS. 35A-D are simplified plan views showing operation of an alternative embodiment of a microfluidic matrix architecture in accordance with the present invention.

DETAILED DESCRIPTION OF THE PRESENT INVENTION

Unless defined otherwise, all technical and scientific terms used herein have the meaning commonly understood by a person skilled in the art to which this invention belongs. The following references provide one of skill with a general definition of many of the terms used in this invention: Singleton et al., *DICTIONARY OF MICROBIOLOGY AND MOLECULAR BIOLOGY* (2d ed. 1994); *THE CAMBRIDGE DICTIONARY OF SCIENCE AND TECHNOLOGY* (Walker ed., 1988); *THE GLOSSARY OF GENETICS*, 5TH ED., R. Rieger et al. (eds.), Springer Verlag (1991); and Hale & Marham, *THE HARPER COLLINS DICTIONARY OF BIOLOGY* (1991). As used herein, the following terms have the meanings ascribed to them unless specified otherwise.

A “flow channel” refers generally to a flow path through which a solution can flow.

The term “valve” unless otherwise indicated refers to a configuration in which a flow channel and a control channel intersect and are separated by an elastomeric membrane that can be deflected into or retracted from the flow channel in response to an actuation force.

The term “elastomer” and “elastomeric” has its general meaning as used in the art. Thus, for example, Allcock et al. (*Contemporary Polymer Chemistry*, 2nd Ed.) describes elastomers in general as polymers existing at a temperature between their glass transition temperature and liquefaction temperature. Elastomeric materials exhibit elastic properties because the polymer chains readily undergo torsional motion to permit uncoiling of the backbone chains in response to a force, with the backbone chains recoiling to assume the prior shape in the absence of the force. In general, elastomers deform when force is applied, but then return to their original shape when the force is removed. The elasticity exhibited by elastomeric materials can be characterized by a Young’s modulus. The elastomeric materials utilized in the microfluidic devices disclosed herein typically have a Young’s modulus of between about 1 Pa-1 TPa, in other instances between about 10 Pa-100 GPa, in still other instances between about 20 Pa-1 GPa, in yet other instances between about 50 Pa-10 MPa, and in certain instances between about 100 Pa-1 MPa. Elastomeric materials having a Young’s modulus outside of these ranges can also be utilized depending upon the needs of a particular application.

Some of the microfluidic devices described herein are fabricated from an elastomeric polymer such as GE RTV 615 (formulation), a vinyl-silane crosslinked (type) silicone elastomer (family). However, the present microfluidic systems are not limited to this one formulation, type or even this family of polymer; rather, nearly any elastomeric polymer is suitable. Given the tremendous diversity of polymer chemistries, precursors, synthetic methods, reaction conditions, and potential additives, there are a large number of possible elastomer systems that can be used to make monolithic elastomeric microvalves and pumps. The choice of materials typically depends upon the particular material properties (e.g., solvent resistance, stiffness, gas permeability, and/or temperature stability) required for the application being conducted. Additional details regarding the type of elastomeric materials that can be used in the manufacture of the components of the microfluidic devices disclosed herein are set forth

in U.S. application Ser. No. 09/605,520, and PCT Application No. 00/17740, both of which are incorporated herein by reference in their entirety.

The terms “nucleic acid,” “polynucleotide,” and “oligonucleotide” are used herein to include a polymeric form of nucleotides of any length, including, but not limited to, ribonucleotides or deoxyribonucleotides. There is no intended distinction in length between these terms. Further, these terms refer only to the primary structure of the molecule. Thus, in certain embodiments these terms can include triple-, double- and single-stranded DNA, as well as triple-, double- and single-stranded RNA. They also include modifications, such as by methylation and/or by capping, and unmodified forms of the polynucleotide. More particularly, the terms “nucleic acid,” “polynucleotide,” and “oligonucleotide,” include polydeoxyribonucleotides (containing 2-deoxy-D-ribose), polyribonucleotides (containing D-ribose), any other type of polynucleotide which is an N- or C-glycoside of a purine or pyrimidine base, and other polymers containing nonnucleotidic backbones, for example, polyamide (e.g., peptide nucleic acids (PNAs)) and polymorpholino (commercially available from the Anti-Virals, Inc., Corvallis, Oreg., as Neugene) polymers, and other synthetic sequence-specific nucleic acid polymers providing that the polymers contain nucleobases in a configuration which allows for base pairing and base stacking, such as is found in DNA and RNA.

A “probe” is an nucleic acid capable of binding to a target nucleic acid of complementary sequence through one or more types of chemical bonds, usually through complementary base pairing, usually through hydrogen bond formation, thus forming a duplex structure. The probe binds or hybridizes to a “probe binding site.” The probe can be labeled with a detectable label to permit facile detection of the probe, particularly once the probe has hybridized to its complementary target. The label attached to the probe can include any of a variety of different labels known in the art that can be detected by chemical or physical means, for example. Suitable labels that can be attached to probes include, but are not limited to, radioisotopes, fluorophores, chromophores, mass labels, electron dense particles, magnetic particles, spin labels, molecules that emit chemiluminescence, electrochemically active molecules, enzymes, cofactors, and enzyme substrates. Probes can vary significantly in size. Some probes are relatively short. Generally, probes are at least 7 to 15 nucleotides in length. Other probes are at least 20, or 40 nucleotides long. Still other probes are somewhat longer, being at least 50, 60, 70, 80, 90 nucleotides long. Yet other probes are longer still, and are at least 100, 150, 200 or more nucleotides long. Probes can be of any specific length that falls within the foregoing ranges as well.

A “primer” is a single-stranded polynucleotide capable of acting as a point of initiation of template-directed DNA synthesis under appropriate conditions (i.e., in the presence of four different nucleoside triphosphates and an agent for polymerization, such as, DNA or RNA polymerase or reverse transcriptase) in an appropriate buffer and at a suitable temperature. The appropriate length of a primer depends on the intended use of the primer but typically is at least 7 nucleotides long and, more typically range from 10 to 30 nucleotides in length. Other primers can be somewhat longer such as 30 to 50 nucleotides long. Short primer molecules generally require cooler temperatures to form sufficiently stable hybrid complexes with the template. A primer need not reflect the exact sequence of the template but must be sufficiently complementary to hybridize with a template. The term “primer site” or “primer binding site” refers to the segment of the target DNA to which a primer hybridizes. The term

“primer pair” means a set of primers including a 5’ “upstream primer” that hybridizes with the complement of the 5’ end of the DNA sequence to be amplified and a 3’ “downstream primer” that hybridizes with the 3’ end of the sequence to be amplified.

A primer that is “perfectly complementary” has a sequence fully complementary across the entire length of the primer and has no mismatches. The primer is typically perfectly complementary to a portion (subsequence) of a target sequence. A “mismatch” refers to a site at which the nucleotide in the primer and the nucleotide in the target nucleic acid with which it is aligned are not complementary. The term “substantially complementary” when used in reference to a primer means that a primer is not perfectly complementary to its target sequence; instead, the primer is only sufficiently complementary to hybridize selectively to its respective strand at the desired primer-binding site.

The term “complementary” means that one nucleic acid is identical to, or hybridizes selectively to, another nucleic acid molecule. Selectivity of hybridization exists when hybridization occurs that is more selective than total lack of specificity. Typically, selective hybridization will occur when there is at least about 55% identity over a stretch of at least 14-25 nucleotides, preferably at least 65%, more preferably at least 75%, and most preferably at least 90%. Preferably, one nucleic acid hybridizes specifically to the other nucleic acid. See M. Kanehisa, *Nucleic Acids Res.* 12:203 (1984).

The term “label” refers to a molecule or an aspect of a molecule that can be detected by physical, chemical, electromagnetic and other related analytical techniques. Examples of detectable labels that can be utilized include, but are not limited to, radioisotopes, fluorophores, chromophores, mass labels, electron dense particles, magnetic particles, spin labels, molecules that emit chemiluminescence, electrochemically active molecules, enzymes, cofactors, enzymes linked to nucleic acid probes and enzyme substrates. The term “detectably labeled” means that an agent has been conjugated with a label or that an agent has some inherent characteristic (e.g., size, shape or color) that allows it to be detected without having to be conjugated to a separate label.

I. Microfabrication Overview

The following discussion relates to formation of microfabricated fluidic devices utilizing elastomer materials, as described generally in U.S. patent application Ser. No. 10/118,466, filed Apr. 5, 2002, Ser. No. 10/265,473 filed Oct. 4, 2002, Ser. No. 10/118,466 filed Apr. 5, 2002, Ser. No. 09/826,585 filed Apr. 6, 2001, Ser. No. 09/724,784 filed Nov. 28, 2000, and Ser. No. 09/605,520, filed Jun. 27, 2000. These previously-filed patent applications are hereby incorporated by reference for all purposes.

1. Methods of Fabricating

Exemplary methods of fabricating the present invention are provided herein. It is to be understood that the present invention is not limited to fabrication by one or the other of these methods. Rather, other suitable methods of fabricating the present microstructures, including modifying the present methods, are also contemplated.

FIGS. 1 to 7B illustrate sequential steps of a first preferred method of fabricating the present microstructure, (which may be used as a pump or valve). FIGS. 8 to 18 illustrate sequential steps of a second preferred method of fabricating the present microstructure, (which also may be used as a pump or valve).

As will be explained, the preferred method of FIGS. 1 to 7B involves using pre-cured elastomer layers which are assembled and bonded. In an alternative method, each layer of elastomer may be cured “in place”. In the following

description “channel” refers to a recess in the elastomeric structure which can contain a flow of fluid or gas.

Referring to FIG. 1, a first micro-machined mold 10 is provided. Micro-machined mold 10 may be fabricated by a number of conventional silicon processing methods, including but not limited to photolithography, ion-milling, and electron beam lithography.

As can be seen, micro-machined mold 10 has a raised line or protrusion 11 extending therealong. A first elastomeric layer 20 is cast on top of mold 10 such that a first recess 21 will be formed in the bottom surface of elastomeric layer 20, (recess 21 corresponding in dimension to protrusion 11), as shown.

As can be seen in FIG. 2, a second micro-machined mold 12 having a raised protrusion 13 extending therealong is also provided. A second elastomeric layer 22 is cast on top of mold 12, as shown, such that a recess 23 will be formed in its bottom surface corresponding to the dimensions of protrusion 13.

As can be seen in the sequential steps illustrated in FIGS. 3 and 4, second elastomeric layer 22 is then removed from mold 12 and placed on top of first elastomeric layer 20. As can be seen, recess 23 extending along the bottom surface of second elastomeric layer 22 will form a flow channel 32.

Referring to FIG. 5, the separate first and second elastomeric layers 20 and 22 (FIG. 4) are then bonded together to form an integrated (i.e.: monolithic) elastomeric structure 24.

As can be seen in the sequential step of FIGS. 6 and 7A, elastomeric structure 24 is then removed from mold 10 and positioned on top of a planar substrate 14. As can be seen in FIGS. 7A and 7B, when elastomeric structure 24 has been sealed at its bottom surface to planar substrate 14, recess 21 will form a flow channel 30.

The present elastomeric structures form a reversible hermetic seal with nearly any smooth planar substrate. An advantage to forming a seal this way is that the elastomeric structures may be peeled up, washed, and re-used. In preferred aspects, planar substrate 14 is glass. A further advantage of using glass is that glass is transparent, allowing optical interrogation of elastomer channels and reservoirs. Alternatively, the elastomeric structure may be bonded onto a flat elastomer layer by the same method as described above, forming a permanent and high-strength bond. This may prove advantageous when higher back pressures are used.

As can be seen in FIGS. 7A and 7B, flow channels 30 and 32 are preferably disposed at an angle to one another with a small membrane 25 of substrate 24 separating the top of flow channel 30 from the bottom of flow channel 32.

In preferred aspects, planar substrate 14 is glass. An advantage of using glass is that the present elastomeric structures may be peeled up, washed and reused. A further advantage of using glass is that optical sensing may be employed. Alternatively, planar substrate 14 may be an elastomer itself, which may prove advantageous when higher back pressures are used.

The method of fabrication just described may be varied to form a structure having a membrane composed of an elastomeric material different than that forming the walls of the channels of the device. This variant fabrication method is illustrated in FIGS. 7C-7G.

Referring to FIG. 7C, a first micro-machined mold 10 is provided. Micro-machined mold 10 has a raised line or protrusion 11 extending therealong. In FIG. 7D, first elastomeric layer 20 is cast on top of first micro-machined mold 10 such that the top of the first elastomeric layer 20 is flush with the top of raised line or protrusion 11. This may be accomplished by carefully controlling the volume of elastomeric material

spun onto mold **10** relative to the known height of raised line **11**. Alternatively, the desired shape could be formed by injection molding.

In FIG. 7E, second micro-machined mold **12** having a raised protrusion **13** extending therealong is also provided. Second elastomeric layer **22** is cast on top of second mold **12** as shown, such that recess **23** is formed in its bottom surface corresponding to the dimensions of protrusion **13**.

In FIG. 7F, second elastomeric layer **22** is removed from mold **12** and placed on top of third elastomeric layer **222**. Second elastomeric layer **22** is bonded to third elastomeric layer **20** to form integral elastomeric block **224** using techniques described in detail below. At this point in the process, recess **23** formerly occupied by raised line **13** will form flow channel **23**.

In FIG. 7G, elastomeric block **224** is placed on top of first micro-machined mold **10** and first elastomeric layer **20**. Elastomeric block and first elastomeric layer **20** are then bonded together to form an integrated (i.e.: monolithic) elastomeric structure **24** having a membrane composed of a separate elastomeric layer **222**.

When elastomeric structure **24** has been sealed at its bottom surface to a planar substrate in the manner described above in connection with FIG. 7A, the recess formerly occupied by raised line **11** will form flow channel **30**.

The variant fabrication method illustrated above in conjunction with FIGS. 7C-7G offers the advantage of permitting the membrane portion to be composed of a separate material than the elastomeric material of the remainder of the structure. This is important because the thickness and elastic properties of the membrane play a key role in operation of the device. Moreover, this method allows the separate elastomer layer to readily be subjected to conditioning prior to incorporation into the elastomer structure. As discussed in detail below, examples of potentially desirable condition include the introduction of magnetic or electrically conducting species to permit actuation of the membrane, and/or the introduction of dopant into the membrane in order to alter its elasticity.

While the above method is illustrated in connection with forming various shaped elastomeric layers formed by replication molding on top of a micromachined mold, the present invention is not limited to this technique. Other techniques could be employed to form the individual layers of shaped elastomeric material that are to be bonded together. For example, a shaped layer of elastomeric material could be formed by laser cutting or injection molding, or by methods utilizing chemical etching and/or sacrificial materials as discussed below in conjunction with the second exemplary method.

An alternative method fabricates a patterned elastomer structure utilizing development of photoresist encapsulated within elastomer material. However, the methods in accordance with the present invention are not limited to utilizing photoresist. Other materials such as metals could also serve as sacrificial materials to be removed selective to the surrounding elastomer material, and the method would remain within the scope of the present invention. For example, gold metal may be etched selective to RTV 615 elastomer utilizing the appropriate chemical mixture.

2. Layer and Channel Dimensions

Microfabricated refers to the size of features of an elastomeric structure fabricated in accordance with an embodiment of the present invention. In general, variation in at least one dimension of microfabricated structures is controlled to the micron level, with at least one dimension being microscopic (i.e. below 1000 μm). Microfabrication typically involves

semiconductor or MEMS fabrication techniques such as photolithography and spincoating that are designed for to produce feature dimensions on the microscopic level, with at least some of the dimension of the microfabricated structure requiring a microscope to reasonably resolve/image the structure.

In preferred aspects, flow channels **30** and **32** preferably have width-to-depth ratios of about 10:1. A non-exclusive list of other ranges of width-to-depth ratios in accordance with embodiments of the present invention is 0.1:1 to 100:1, more preferably 1:1 to 50:1, more preferably 2:1 to 20:1, and most preferably 3:1 to 15:1. In an exemplary aspect, flow channels **30**, **32**, **60** and **62** have widths of about 1 to 1000 microns. A non-exclusive list of other ranges of widths of flow channels in accordance with embodiments of the present invention is 0.01 to 1000 microns, more preferably 0.05 to 1000 microns, more preferably 0.2 to 500 microns, more preferably 1 to 250 microns, and most preferably 10 to 200 microns. Exemplary channel widths include 0.1 μm , 1 μm , 2 μm , 5 μm , 10 μm , 20 μm , 30 μm , 40 μm , 50 μm , 60 μm , 70 μm , 80 μm , 90 μm , 100 μm , 110 μm , 120 μm , 130 μm , 140 μm , 150 μm , 160 μm , 170 μm , 180 μm , 190 μm , 200 μm , 210 μm , 220 μm , 230 μm , 240 μm , and 250 μm .

Flow channels **30** and **32** have depths of about 1 to 100 microns. A non-exclusive list of other ranges of depths of flow channels in accordance with embodiments of the present invention is 0.01 to 1000 microns, more preferably 0.05 to 500 microns, more preferably 0.2 to 250 microns, and more preferably 1 to 100 microns, more preferably 2 to 20 microns, and most preferably 5 to 10 microns. Exemplary channel depths include including 0.01 μm , 0.02 μm , 0.05 μm , 0.1 μm , 0.2 μm , 0.5 μm , 1 μm , 2 μm , 3 μm , 4 μm , 5 μm , 7.5 μm , 10 μm , 12.5 μm , 15 μm , 17.5 μm , 20 μm , 22.5 μm , 25 μm , 30 μm , 40 μm , 50 μm , 75 μm , 100 μm , 150 μm , 200 μm , and 250 μm .

The flow channels are not limited to these specific dimension ranges and examples given above, and may vary in width in order to affect the magnitude of force required to deflect the membrane as discussed at length below in conjunction with FIG. 27. For example, extremely narrow flow channels having a width on the order of 0.01 μm may be useful in optical and other applications, as discussed in detail below. Elastomeric structures which include portions having channels of even greater width than described above are also contemplated by the present invention, and examples of applications of utilizing such wider flow channels include fluid reservoir and mixing channel structures.

The Elastomeric layers may be cast thick for mechanical stability. In an exemplary embodiment, elastomeric layer **22** of FIG. 1 is 50 microns to several centimeters thick, and more preferably approximately 4 mm thick. A non-exclusive list of ranges of thickness of the elastomer layer in accordance with other embodiments of the present invention is between about 0.1 micron to 10 cm, 1 micron to 5 cm, 10 microns to 2 cm, 100 microns to 10 mm.

Accordingly, membrane **25** of FIG. 7B separating flow channels **30** and **32** has a typical thickness of between about 0.01 and 1000 microns, more preferably 0.05 to 500 microns, more preferably 0.2 to 250, more preferably 1 to 100 microns, more preferably 2 to 50 microns, and most preferably 5 to 40 microns. As such, the thickness of elastomeric layer **22** is about 100 times the thickness of elastomeric layer **20**. Exemplary membrane thicknesses include 0.01 μm , 0.02 μm , 0.03 μm , 0.05 μm , 0.1 μm , 0.2 μm , 0.3 μm , 0.5 μm , 1 μm , 2 μm , 3 μm , 5 μm , 7.5 μm , 10 μm , 12.5 μm , 15 μm , 17.5 μm , 20 μm , 22.5 μm , 25 μm , 30 μm , 40 μm , 50 μm , 75 μm , 100 μm , 150 μm , 200 μm , 250 μm , 300 μm , 400 μm , 500 μm , 750 μm , and 1000 μm .

3. Soft Lithographic Bonding

Preferably, elastomeric layers are bonded together chemically, using chemistry that is intrinsic to the polymers comprising the patterned elastomer layers. Most preferably, the bonding comprises two component “addition cure” bonding.

In a preferred aspect, the various layers of elastomer are bound together in a heterogeneous bonding in which the layers have a different chemistry. Alternatively, a homogeneous bonding may be used in which all layers would be of the same chemistry. Thirdly, the respective elastomer layers may optionally be glued together by an adhesive instead. In a fourth aspect, the elastomeric layers may be thermoset elastomers bonded together by heating.

In one aspect of homogeneous bonding, the elastomeric layers are composed of the same elastomer material, with the same chemical entity in one layer reacting with the same chemical entity in the other layer to bond the layers together. In one embodiment, bonding between polymer chains of like elastomer layers may result from activation of a crosslinking agent due to light, heat, or chemical reaction with a separate chemical species.

Alternatively in a heterogeneous aspect, the elastomeric layers are composed of different elastomeric materials, with a first chemical entity in one layer reacting with a second chemical entity in another layer. In one exemplary heterogeneous aspect, the bonding process used to bind respective elastomeric layers together may comprise bonding together two layers of RTV 615 silicone. RTV 615 silicone is a two-part addition-cure silicone rubber. Part A contains vinyl groups and catalyst; part B contains silicon hydride (Si—H) groups. The conventional ratio for RTV 615 is 10A:1B. For bonding, one layer may be made with 30A:1B (i.e. excess vinyl groups) and the other with 3A:1B (i.e. excess Si—H groups). Each layer is cured separately. When the two layers are brought into contact and heated at elevated temperature, they bond irreversibly forming a monolithic elastomeric substrate.

In an exemplary aspect of the present invention, elastomeric structures are formed utilizing Sylgard 182, 184 or 186, or aliphatic urethane diacrylates such as (but not limited to) Ebecryl 270 or Irr 245 from UCB Chemical.

In one embodiment in accordance with the present invention, two-layer elastomeric structures were fabricated from pure acrylated Urethane Ebe 270. A thin bottom layer was spin coated at 8000 rpm for 15 seconds at 170° C. The top and bottom layers were initially cured under ultraviolet light for 10 minutes under nitrogen utilizing a Model ELC 500 device manufactured by Electrolite corporation. The assembled layers were then cured for an additional 30 minutes. Reaction was catalyzed by a 0.5% vol/vol mixture of Irgacure 500 manufactured by Ciba-Geigy Chemicals. The resulting elastomeric material exhibited moderate elasticity and adhesion to glass.

In another embodiment in accordance with the present invention, two-layer elastomeric structures were fabricated from a combination of 25% Ebe 270/50% Irr245/25% isopropyl alcohol for a thin bottom layer, and pure acrylated Urethane Ebe 270 as a top layer. The thin bottom layer was initially cured for 5 min, and the top layer initially cured for 10 minutes, under ultraviolet light under nitrogen utilizing a Model ELC 500 device manufactured by Electrolite corporation. The assembled layers were then cured for an additional 30 minutes. Reaction was catalyzed by a 0.5% vol/vol mixture of Irgacure 500 manufactured by Ciba-Geigy Chemicals. The resulting elastomeric material exhibited moderate elasticity and adhered to glass.

Alternatively, other bonding methods may be used, including activating the elastomer surface, for example by plasma exposure, so that the elastomer layers/substrate will bond when placed in contact. For example, one possible approach to bonding together elastomer layers composed of the same material is set forth by Duffy et al, “Rapid Prototyping of Microfluidic Systems in Poly (dimethylsiloxane)”, *Analytical Chemistry* (1998), 70, 4974-4984, incorporated herein by reference. This paper discusses that exposing polydimethylsiloxane (PDMS) layers to oxygen plasma causes oxidation of the surface, with irreversible bonding occurring when the two oxidized layers are placed into contact.

Yet another approach to bonding together successive layers of elastomer is to utilize the adhesive properties of uncured elastomer. Specifically, a thin layer of uncured elastomer such as RTV 615 is applied on top of a first cured elastomeric layer. Next, a second cured elastomeric layer is placed on top of the uncured elastomeric layer. The thin middle layer of uncured elastomer is then cured to produce a monolithic elastomeric structure. Alternatively, uncured elastomer can be applied to the bottom of a first cured elastomer layer, with the first cured elastomer layer placed on top of a second cured elastomer layer. Curing the middle thin elastomer layer again results in formation of a monolithic elastomeric structure.

Where encapsulation of sacrificial layers is employed to fabricate the elastomer structure, bonding of successive elastomeric layers may be accomplished by pouring uncured elastomer over a previously cured elastomeric layer and any sacrificial material patterned thereupon. Bonding between elastomer layers occurs due to interpenetration and reaction of the polymer chains of an uncured elastomer layer with the polymer chains of a cured elastomer layer. Subsequent curing of the elastomeric layer will create a bond between the elastomeric layers and create a monolithic elastomeric structure.

Referring to the first method of FIGS. 1 to 7B, first elastomeric layer 20 may be created by spin-coating an RTV mixture on microfabricated mold 10 at 2000 rpm’s for 30 seconds yielding a thickness of approximately 40 microns. Second elastomeric layer 22 may be created by spin-coating an RTV mixture on microfabricated mold 12. Both layers 20 and 22 may be separately baked or cured at about 80° C. for 1.5 hours. The second elastomeric layer 22 may be bonded onto first elastomeric layer 20 at about 80° C. for about 1.5 hours.

Micromachined molds 10 and 12 may be patterned photoresist on silicon wafers. In an exemplary aspect, a Shipley SJR 5740 photoresist was spun at 2000 rpm patterned with a high resolution transparency film as a mask and then developed yielding an inverse channel of approximately 10 microns in height. When baked at approximately 200° C. for about 30 minutes, the photoresist reflows and the inverse channels become rounded. In preferred aspects, the molds may be treated with trimethylchlorosilane (TMCS) vapor for about a minute before each use in order to prevent adhesion of silicone rubber.

4. Suitable Elastomeric Materials

Allcock et al, *Contemporary Polymer Chemistry*, 2nd Ed. describes elastomers in general as polymers existing at a temperature between their glass transition temperature and liquefaction temperature. Elastomeric materials exhibit elastic properties because the polymer chains readily undergo torsional motion to permit uncoiling of the backbone chains in response to a force, with the backbone chains recoiling to assume the prior shape in the absence of the force. In general, elastomers deform when force is applied, but then return to their original shape when the force is removed. The elasticity exhibited by elastomeric materials may be characterized by a Young’s modulus. Elastomeric materials having a Young’s

modulus of between about 1 Pa-1 TPa, more preferably between about 10 Pa-100 GPa, more preferably between about 20 Pa-1 GPa, more preferably between about 50 Pa-10 MPa, and more preferably between about 100 Pa-1 MPa are useful in accordance with the present invention, although elastomeric materials having a Young's modulus outside of these ranges could also be utilized depending upon the needs of a particular application.

The systems of the present invention may be fabricated from a wide variety of elastomers. In an exemplary aspect, the elastomeric layers may preferably be fabricated from silicone rubber. However, other suitable elastomers may also be used.

In an exemplary aspect of the present invention, the present systems are fabricated from an elastomeric polymer such as GE RTV 615 (formulation), a vinyl-silane crosslinked (type) silicone elastomer (family). However, the present systems are not limited to this one formulation, type or even this family of polymer; rather, nearly any elastomeric polymer is suitable. An important requirement for the preferred method of fabrication of the present microvalves is the ability to bond multiple layers of elastomers together. In the case of multilayer soft lithography, layers of elastomer are cured separately and then bonded together. This scheme requires that cured layers possess sufficient reactivity to bond together. Either the layers may be of the same type, and are capable of bonding to themselves, or they may be of two different types, and are capable of bonding to each other. Other possibilities include the use of an adhesive between layers and the use of thermoset elastomers.

Given the tremendous diversity of polymer chemistries, precursors, synthetic methods, reaction conditions, and potential additives, there are a huge number of possible elastomer systems that could be used to make monolithic elastomeric microvalves and pumps. Variations in the materials used will most likely be driven by the need for particular material properties, i.e. solvent resistance, stiffness, gas permeability, or temperature stability.

There are many, many types of elastomeric polymers. A brief description of the most common classes of elastomers is presented here, with the intent of showing that even with relatively "standard" polymers, many possibilities for bonding exist. Common elastomeric polymers include polyisoprene, polybutadiene, polychloroprene, polyisobutylene, poly(styrene-butadiene-styrene), the polyurethanes, and silicones.

Polyisoprene, Polybutadiene, Polychloroprene:

Polyisoprene, polybutadiene, and polychloroprene are all polymerized from diene monomers, and therefore have one double bond per monomer when polymerized. This double bond allows the polymers to be converted to elastomers by vulcanization (essentially, sulfur is used to form crosslinks between the double bonds by heating). This would easily allow homogeneous multilayer soft lithography by incomplete vulcanization of the layers to be bonded; photoresist encapsulation would be possible by a similar mechanism.

Polyisobutylene:

Pure Polyisobutylene has no double bonds, but is crosslinked to use as an elastomer by including a small amount (~1%) of isoprene in the polymerization. The isoprene monomers give pendant double bonds on the Polyisobutylene backbone, which may then be vulcanized as above.

Poly(styrene-butadiene-styrene):

Poly(styrene-butadiene-styrene) is produced by living anionic polymerization (that is, there is no natural chain-terminating step in the reaction), so "live" polymer ends can exist in the cured polymer. This makes it a natural candidate

for the present photoresist encapsulation system (where there will be plenty of unreacted monomer in the liquid layer poured on top of the cured layer). Incomplete curing would allow homogeneous multilayer soft lithography (A to A bonding). The chemistry also facilitates making one layer with extra butadiene ("A") and coupling agent and the other layer ("B") with a butadiene deficit (for heterogeneous multilayer soft lithography). SBS is a "thermoset elastomer", meaning that above a certain temperature it melts and becomes plastic (as opposed to elastic); reducing the temperature yields the elastomer again. Thus, layers can be bonded together by heating.

Polyurethanes:

Polyurethanes are produced from di-isocyanates (A-A) and di-alcohols or di-amines (B-B); since there are a large variety of di-isocyanates and di-alcohols/amines, the number of different types of polyurethanes is huge. The A vs. B nature of the polymers, however, would make them useful for heterogeneous multilayer soft lithography just as RTV 615 is: by using excess A-A in one layer and excess B-B in the other layer.

Silicones:

Silicone polymers probably have the greatest structural variety, and almost certainly have the greatest number of commercially available formulations. The vinyl-to-(Si-H) crosslinking of RTV 615 (which allows both heterogeneous multilayer soft lithography and photoresist encapsulation) has already been discussed, but this is only one of several crosslinking methods used in silicone polymer chemistry.

5. Operation of Device

FIGS. 7B and 7H together show the closing of a first flow channel by pressurizing a second flow channel, with FIG. 7B (a front sectional view cutting through flow channel 32 in corresponding FIG. 7A), showing an open first flow channel 30; with FIG. 7H showing first flow channel 30 closed by pressurization of the second flow channel 32.

Referring to FIG. 7B, first flow channel 30 and second flow channel 32 are shown. Membrane 25 separates the flow channels, forming the top of first flow channel 30 and the bottom of second flow channel 32. As can be seen, flow channel 30 is "open".

As can be seen in FIG. 7H, pressurization of flow channel 32 (either by gas or liquid introduced therein) causes membrane 25 to deflect downward, thereby pinching off flow F passing through flow channel 30. Accordingly, by varying the pressure in channel 32, a linearly actuatable valving system is provided such that flow channel 30 can be opened or closed by moving membrane 25 as desired. (For illustration purposes only, channel 30 in FIG. 7G is shown in a "mostly closed" position, rather than a "fully closed" position).

Since such valves are actuated by moving the roof of the channels themselves (i.e.: moving membrane 25) valves and pumps produced by this technique have a truly zero dead volume, and switching valves made by this technique have a dead volume approximately equal to the active volume of the valve, for example about $100 \times 100 \times 10 \mu\text{m} = 100 \text{ pL}$. Such dead volumes and areas consumed by the moving membrane are approximately two orders of magnitude smaller than known conventional microvalves. Smaller and larger valves and switching valves are contemplated in the present invention, and a non-exclusive list of ranges of dead volume includes 1 aL to 1 uL, 100 aL to 100 mL, 1 fL to 10 mL, 100 fL to 1 mL, and 1 pL to 100 pL.

The extremely small volumes capable of being delivered by pumps and valves in accordance with the present invention represent a substantial advantage. Specifically, the smallest known volumes of fluid capable of being manually metered is

around 0.1 μl . The smallest known volumes capable of being metered by automated systems is about ten-times larger (1 μl). Utilizing pumps and valves in accordance with the present invention, volumes of liquid of 10 nl or smaller can routinely be metered and dispensed. The accurate metering of extremely small volumes of fluid enabled by the present invention would be extremely valuable in a large number of biological applications, including diagnostic tests and assays.

Equation 1 represents a highly simplified mathematical model of deflection of a rectangular, linear, elastic, isotropic plate of uniform thickness by an applied pressure:

$$w=(BPb^4)/(Eh^3), \quad (1)$$

where:

w=deflection of plate;

B=shape coefficient (dependent upon length vs. width and support of edges of plate);

P=applied pressure;

b=plate width

E=Young's modulus; and

h=plate thickness.

Thus even in this extremely simplified expression, deflection of an elastomeric membrane in response to a pressure will be a function of: the length, width, and thickness of the membrane, the flexibility of the membrane (Young's modulus), and the applied actuation force. Because each of these parameters will vary widely depending upon the actual dimensions and physical composition of a particular elastomeric device in accordance with the present invention, a wide range of membrane thicknesses and elasticity's, channel widths, and actuation forces are contemplated by the present invention.

It should be understood that the formula just presented is only an approximation, since in general the membrane does not have uniform thickness, the membrane thickness is not necessarily small compared to the length and width, and the deflection is not necessarily small compared to length, width, or thickness of the membrane. Nevertheless, the equation serves as a useful guide for adjusting variable parameters to achieve a desired response of deflection versus applied force.

FIGS. 8A and 8B illustrate valve opening vs. applied pressure for a 100 μm wide first flow channel 30 and a 50 μm wide second flow channel 32. The membrane of this device was formed by a layer of General Electric Silicones RTV 615 having a thickness of approximately 30 μm and a Young's modulus of approximately 750 kPa. FIGS. 21a and 21b show the extent of opening of the valve to be substantially linear over most of the range of applied pressures.

Air pressure was applied to actuate the membrane of the device through a 10 cm long piece of plastic tubing having an outer diameter of 0.025" connected to a 25 mm piece of stainless steel hypodermic tubing with an outer diameter of 0.025" and an inner diameter of 0.013". This tubing was placed into contact with the control channel by insertion into the elastomeric block in a direction normal to the control channel. Air pressure was applied to the hypodermic tubing from an external LHDA miniature solenoid valve manufactured by Lee Co.

While control of the flow of material through the device has so far been described utilizing applied gas pressure, other fluids could be used.

For example, air is compressible, and thus experiences some finite delay between the time of application of pressure by the external solenoid valve and the time that this pressure is experienced by the membrane. In an alternative embodiment of the present invention, pressure could be applied from an external source to a noncompressible fluid such as water or

hydraulic oils, resulting in a near-instantaneous transfer of applied pressure to the membrane. However, if the displaced volume of the valve is large or the control channel is narrow, higher viscosity of a control fluid may contribute to delay in actuation. The optimal medium for transferring pressure will therefore depend upon the particular application and device configuration, and both gaseous and liquid media are contemplated by the invention.

While external applied pressure as described above has been applied by a pump/tank system through a pressure regulator and external miniature valve, other methods of applying external pressure are also contemplated in the present invention, including gas tanks, compressors, piston systems, and columns of liquid. Also contemplated is the use of naturally occurring pressure sources such as may be found inside living organisms, such as blood pressure, gastric pressure, the pressure present in the cerebrospinal fluid, pressure present in the intra-ocular space, and the pressure exerted by muscles during normal flexure. Other methods of regulating external pressure are also contemplated, such as miniature valves, pumps, macroscopic peristaltic pumps, pinch valves, and other types of fluid regulating equipment such as is known in the art.

As can be seen, the response of valves in accordance with embodiments of the present invention have been experimentally shown to be almost perfectly linear over a large portion of its range of travel, with minimal hysteresis. Accordingly, the present valves are ideally suited for microfluidic metering and fluid control. The linearity of the valve response demonstrates that the individual valves are well modeled as Hooke's Law springs. Furthermore, high pressures in the flow channel (i.e.: back pressure) can be countered simply by increasing the actuation pressure. Experimentally, the present inventors have achieved valve closure at back pressures of 70 kPa, but higher pressures are also contemplated. The following is a nonexclusive list of pressure ranges encompassed by the present invention: 10 Pa-25 MPa; 100 Pa-10 Mpa, 1 kPa-1 MPa, 1 kPa-300 kPa, 5 kPa-200 kPa, and 15 kPa-100 kPa.

While valves and pumps do not require linear actuation to open and close, linear response does allow valves to more easily be used as metering devices. In one embodiment of the invention, the opening of the valve is used to control flow rate by being partially actuated to a known degree of closure. Linear valve actuation makes it easier to determine the amount of actuation force required to close the valve to a desired degree of closure. Another benefit of linear actuation is that the force required for valve actuation may be easily determined from the pressure in the flow channel. If actuation is linear, increased pressure in the flow channel may be countered by adding the same pressure (force per unit area) to the actuated portion of the valve.

Linearity of a valve depends on the structure, composition, and method of actuation of the valve structure. Furthermore, whether linearity is a desirable characteristic in a valve depends on the application. Therefore, both linearly and non-linearly actuatable valves are contemplated in the present invention, and the pressure ranges over which a valve is linearly actuatable will vary with the specific embodiment.

FIG. 9 illustrates time response (i.e.: closure of valve as a function of time in response to a change in applied pressure) of a 100 μm ×100 μm ×10 μm RTV microvalve with 10-cm-long air tubing connected from the chip to a pneumatic valve as described above.

Two periods of digital control signal, actual air pressure at the end of the tubing and valve opening are shown in FIG. 9. The pressure applied on the control line is 100 kPa, which is substantially higher than the ~40 kPa required to close the

valve. Thus, when closing, the valve is pushed closed with a pressure 60 kPa greater than required. When opening, however, the valve is driven back to its rest position only by its own spring force (≤ 40 kPa). Thus, τ_{close} is expected to be smaller than τ_{open} . There is also a lag between the control signal and control pressure response, due to the limitations of the miniature valve used to control the pressure. Calling such lags t and the $1/e$ time constants τ , the values are: $\tau_{\text{open}}=3.63$ ms, $\tau_{\text{open}}=1.88$ ms, $\tau_{\text{close}}=2.15$ ms, $\tau_{\text{close}}=0.51$ ms. If 3τ each are allowed for opening and closing, the valve runs comfortably at 75 Hz when filled with aqueous solution.

If one used another actuation method which did not suffer from opening and closing lag, this valve would run at ~ 375 Hz. Note also that the spring constant can be adjusted by changing the membrane thickness; this allows optimization for either fast opening or fast closing. The spring constant could also be adjusted by changing the elasticity (Young's modulus) of the membrane, as is possible by introducing dopant into the membrane or by utilizing a different elastomeric material to serve as the membrane (described above in conjunction with FIGS. 7C-7H.)

When experimentally measuring the valve properties as illustrated in FIG. 9 the valve opening was measured by fluorescence. In these experiments, the flow channel was filled with a solution of fluorescein isothiocyanate (FITC) in buffer ($\text{pH} \geq 8$) and the fluorescence of a square area occupying the center $\sim 1/3$ rd of the channel is monitored on an epifluorescence microscope with a photomultiplier tube with a 10 kHz bandwidth. The pressure was monitored with a Wheatstone-bridge pressure sensor (SenSym SCC15GD2) pressurized simultaneously with the control line through nearly identical pneumatic connections.

6. Flow Channel Cross Sections

The flow channels of the present invention may optionally be designed with different cross sectional sizes and shapes, offering different advantages, depending upon their desired application. For example, the cross sectional shape of the lower flow channel may have a curved upper surface, either along its entire length or in the region disposed under an upper cross channel). Such a curved upper surface facilitates valve sealing, as follows.

Referring to FIG. 10, a cross sectional view (similar to that of FIG. 7B) through flow channels 30 and 32 is shown. As can be seen, flow channel 30 is rectangular in cross sectional shape. In an alternate preferred aspect of the invention, as shown in FIG. 10, the cross-section of a flow channel 30 instead has an upper curved surface.

Referring first to FIG. 10, when flow channel 32 is pressurized, the membrane portion 25 of elastomeric block 24 separating flow channels 30 and 32 will move downwardly to the successive positions shown by the dotted lines 25A, 25B, 25C, 25D, and 25E. As can be seen, incomplete sealing may possibly result at the edges of flow channel 30 adjacent planar substrate 14.

In the alternate preferred embodiment of FIG. 11, flow channel 30a has a curved upper wall 25A. When flow channel 32 is pressurized, membrane portion 25 will move downwardly to the successive positions shown by dotted lines 25A2, 25A3, 25A4 and 25A5, with edge portions of the membrane moving first into the flow channel, followed by top membrane portions. An advantage of having such a curved upper surface at membrane 25A is that a more complete seal will be provided when flow channel 32 is pressurized. Specifically, the upper wall of the flow channel 30 will provide a continuous contacting edge against planar substrate 14, thereby avoiding the "island" of contact seen between wall 25 and the bottom of flow channel 30 in FIG. 10.

Another advantage of having a curved upper flow channel surface at membrane 25A is that the membrane can more readily conform to the shape and volume of the flow channel in response to actuation. Specifically, where a rectangular flow channel is employed, the entire perimeter ($2 \times$ flow channel height, plus the flow channel width) must be forced into the flow channel. However where an arched flow channel is used, a smaller perimeter of material (only the semi-circular arched portion) must be forced into the channel. In this manner, the membrane requires less change in perimeter for actuation and is therefore more responsive to an applied actuation force to block the flow channel.

In an alternate aspect, (not illustrated), the bottom of flow channel 30 is rounded such that its curved surface mates with the curved upper wall 25A as seen in FIG. 20 described above.

In summary, the actual conformational change experienced by the membrane upon actuation will depend upon the configuration of the particular elastomeric structure. Specifically, the conformational change will depend upon the length, width, and thickness profile of the membrane, its attachment to the remainder of the structure, and the height, width, and shape of the flow and control channels and the material properties of the elastomer used. The conformational change may also depend upon the method of actuation, as actuation of the membrane in response to an applied pressure will vary somewhat from actuation in response to a magnetic or electrostatic force.

Moreover, the desired conformational change in the membrane will also vary depending upon the particular application for the elastomeric structure. In the simplest embodiments described above, the valve may either be open or closed, with metering to control the degree of closure of the valve. In other embodiments however, it may be desirable to alter the shape of the membrane and/or the flow channel in order to achieve more complex flow regulation. For instance, the flow channel could be provided with raised protrusions beneath the membrane portion, such that upon actuation the membrane shuts off only a percentage of the flow through the flow channel, with the percentage of flow blocked insensitive to the applied actuation force.

Many membrane thickness profiles and flow channel cross-sections are contemplated by the present invention, including rectangular, trapezoidal, circular, ellipsoidal, parabolic, hyperbolic, and polygonal, as well as sections of the above shapes. More complex cross-sectional shapes, such as the embodiment with protrusions discussed immediately above or an embodiment having concavities in the flow channel, are also contemplated by the present invention.

In addition, while the invention is described primarily above in conjunction with an embodiment wherein the walls and ceiling of the flow channel are formed from elastomer, and the floor of the channel is formed from an underlying substrate, the present invention is not limited to this particular orientation. Walls and floors of channels could also be formed in the underlying substrate, with only the ceiling of the flow channel constructed from elastomer. This elastomer flow channel ceiling would project downward into the channel in response to an applied actuation force, thereby controlling the flow of material through the flow channel. In general, monolithic elastomer structures as described elsewhere in the instant application are preferred for microfluidic applications. However, it may be useful to employ channels formed in the substrate where such an arrangement provides advantages. For instance, a substrate including optical waveguides could be constructed so that the optical waveguides direct light specifically to the side of a microfluidic channel.

7. Networked Systems

FIGS. 12A and 12B show a views of a single on/off valve, identical to the systems set forth above, (for example in FIG. 7A). FIGS. 13A and 13B shows a peristaltic pumping system comprised of a plurality of the single addressable on/off valves as seen in FIG. 12, but networked together. FIG. 14 is a graph showing experimentally achieved pumping rates vs. frequency for the peristaltic pumping system of FIG. 13. FIGS. 15A and 15B show a schematic view of a plurality of flow channels which are controllable by a single control line. This system is also comprised of a plurality of the single addressable on/off valves of FIG. 12, multiplexed together, but in a different arrangement than that of FIG. 12. FIG. 16 is a schematic illustration of a multiplexing system adapted to permit fluid flow through selected channels, comprised of a plurality of the single on/off valves of FIG. 12, joined or networked together.

Referring first to FIGS. 12A and 12B, a schematic of flow channels 30 and 32 is shown. Flow channel 30 preferably has a fluid (or gas) flow F passing therethrough. Flow channel 32, (which crosses over flow channel 30, as was already explained herein), is pressurized such that membrane 25 separating the flow channels may be depressed into the path of flow channel 30, shutting off the passage of flow F there-through, as has been explained. As such, "flow channel" 32 can also be referred to as a "control line" which actuates a single valve in flow channel 30. In FIGS. 12 to 15, a plurality of such addressable valves are joined or networked together in various arrangements to produce pumps, capable of peristaltic pumping, and other fluidic logic applications.

Referring to FIGS. 13A and 13B, a system for peristaltic pumping is provided, as follows. A flow channel 30 has a plurality of generally parallel flow channels (i.e.: control lines) 32A, 32B and 32C passing thereover. By pressurizing control line 32A, flow F through flow channel 30 is shut off under membrane 25A at the intersection of control line 32A and flow channel 30. Similarly, (but not shown), by pressurizing control line 32B, flow F through flow channel 30 is shut off under membrane 25B at the intersection of control line 32B and flow channel 30, etc.

Each of control lines 32A, 32B, and 32C is separately addressable. Therefore, peristalsis may be actuated by the pattern of actuating 32A and 32C together, followed by 32A, followed by 32A and 32B together, followed by 32B, followed by 32B and C together, etc. This corresponds to a successive "101, 100, 110, 010, 011, 001" pattern, where "0" indicates "valve open" and "1" indicates "valve closed." This peristaltic pattern is also known as a 120° pattern (referring to the phase angle of actuation between three valves). Other peristaltic patterns are equally possible, including 60° and 90° patterns.

In experiments performed by the inventors, a pumping rate of 2.35 mL/s was measured by measuring the distance traveled by a column of water in thin (0.5 mm i.d.) tubing; with 100×100×10 µm valves under an actuation pressure of 40 kPa. The pumping rate increased with actuation frequency until approximately 75 Hz, and then was nearly constant until above 200 Hz. The valves and pumps are also quite durable and the elastomer membrane, control channels, or bond have never been observed to fail. In experiments performed by the inventors, none of the valves in the peristaltic pump described herein show any sign of wear or fatigue after more than 4 million actuations. In addition to their durability, they are also gentle. A solution of *E. Coli* pumped through a channel and tested for viability showed a 94% survival rate.

FIG. 14 is a graph showing experimentally achieved pumping rates vs. frequency for the peristaltic pumping system of FIG. 13.

FIGS. 15A and 15B illustrates another way of assembling a plurality of the addressable valves of FIG. 12. Specifically, a plurality of parallel flow channels 30A, 30B, and 30C are provided. Flow channel (i.e.: control line) 32 passes thereover across flow channels 30A, 30B, and 30C. Pressurization of control line 32 simultaneously shuts off flows F1, F2 and F3 by depressing membranes 25A, 25B, and 25C located at the intersections of control line 32 and flow channels 30A, 30B, and 30C.

FIG. 16 is a schematic illustration of a multiplexing system adapted to selectively permit fluid to flow through selected channels, as follows. The downward deflection of membranes separating the respective flow channels from a control line passing thereabove (for example, membranes 25A, 25B, and 25C in FIGS. 15A and 15B) depends strongly upon the membrane dimensions. Accordingly, by varying the widths of flow channel control line 32 in FIGS. 15A and 15B, it is possible to have a control line pass over multiple flow channels, yet only actuate (i.e.: seal) desired flow channels. FIG. 16 illustrates a schematic of such a system, as follows.

A plurality of parallel flow channels 30A, 30B, 30C, 30D, 30E and 30F are positioned under a plurality of parallel control lines 32A, 32B, 32C and 32D. Control channels 32A, 32B, 32C and 32D are adapted to shut off fluid flows F1, F2, F3, F4, F5 and F6 passing through parallel flow channels 30A, 30B, 30C, 30D, 30E and 30F using any of the valving systems described above, with the following modification.

Each of control lines 32A, 32B, 32C and 32D have both wide and narrow portions. For example, control line 32A is wide in locations disposed over flow channels 30A, 30C and 30E. Similarly, control line 32B is wide in locations disposed over flow channels 30B, 30D and 30F, and control line 32C is wide in locations disposed over flow channels 30A, 30B, 30E and 30F.

At the locations where the respective control line is wide, its pressurization will cause the membrane (25) separating the flow channel and the control line to depress significantly into the flow channel, thereby blocking the flow passage there-through. Conversely, in the locations where the respective control line is narrow, membrane (25) will also be narrow. Accordingly, the same degree of pressurization will not result in membrane (25) becoming depressed into the flow channel (30). Therefore, fluid passage thereunder will not be blocked.

For example, when control line 32A is pressurized, it will block flows F1, F3 and F5 in flow channels 30A, 30C and 30E. Similarly, when control line 32C is pressurized, it will block flows F1, F2, F5 and F6 in flow channels 30A, 30B, 30E and 30F. As can be appreciated, more than one control line can be actuated at the same time. For example, control lines 32A and 32C can be pressurized simultaneously to block all fluid flow except F4 (with 32A blocking F1, F3 and F5; and 32C blocking F1, F2, F5 and F6).

By selectively pressurizing different control lines (32) both together and in various sequences, a great degree of fluid flow control can be achieved. Moreover, by extending the present system to more than six parallel flow channels (30) and more than four parallel control lines (32), and by varying the positioning of the wide and narrow regions of the control lines, very complex fluid flow control systems may be fabricated. A property of such systems is that it is possible to turn on any one flow channel out of n flow channels with only $2(\log 2n)$ control lines.

8. Switchable Flow Arrays

In yet another novel embodiment, fluid passage can be selectively directed to flow in either of two perpendicular directions. An example of such a “switchable flow array” system is provided in FIGS. 17A to 17D. FIG. 17A shows a bottom view of a first layer of elastomer **90**, (or any other suitable substrate), having a bottom surface with a pattern of recesses forming a flow channel grid defined by an array of solid posts **92**, each having flow channels passing there-around.

In preferred aspects, an additional layer of elastomer is bound to the top surface of layer **90** such that fluid flow can be selectively directed to move either in direction **F1**, or perpendicular direction **F2**. FIG. 17B is a bottom view of the bottom surface of the second layer of elastomer **95** showing recesses formed in the shape of alternating “vertical” control lines **96** and “horizontal” control lines **94**. “Vertical” control lines **96** have the same width therealong, whereas “horizontal” control lines **94** have alternating wide and narrow portions, as shown.

Elastomeric layer **95** is positioned over top of elastomeric layer **90** such that “vertical” control lines **96** are positioned over posts **92** as shown in FIG. 17C and “horizontal” control lines **94** are positioned with their wide portions between posts **92**, as shown in FIG. 17D.

As can be seen in FIG. 17C, when “vertical” control lines **96** are pressurized, the membrane of the integrated structure formed by the elastomeric layer initially positioned between layers **90** and **95** in regions **98** will be deflected downwardly over the array of flow channels such that flow is only able to pass in flow direction **F2** (i.e.: vertically), as shown.

As can be seen in FIG. 17D, when “horizontal” control lines **94** are pressurized, the membrane of the integrated structure formed by the elastomeric layer initially positioned between layers **90** and **95** in regions **99** will be deflected downwardly over the array of flow channels, (but only in the regions where they are widest), such that flow is only able to pass in flow direction **F1** (i.e.: horizontally), as shown.

The design illustrated in FIGS. 17A-D allows a switchable flow array to be constructed from only two elastomeric layers, with no vertical vias passing between control lines in different elastomeric layers required. If all vertical flow control lines **94** are connected, they may be pressurized from one input. The same is true for all horizontal flow control lines **96**.

9. Cell Pen

In yet a further application of the present invention, an elastomeric structure can be utilized to manipulate organisms or other biological material. FIGS. 18A-18D show plan views of one embodiment of a cell pen structure in accordance with the present invention.

Cell pen array **4400** features an array of orthogonally-oriented flow channels **4402**, with an enlarged “pen” structure **4404** at the intersection of alternating flow channels. Valve **4406** is positioned at the entrance and exit of each pen structure **4404**. Peristaltic pump structures **4408** are positioned on each horizontal flow channel and on the vertical flow channels lacking a cell pen structure.

Cell pen array **4400** of FIG. 18A has been loaded with cells A-H that have been previously sorted. FIGS. 18B-18C show the accessing and removal of individually stored cell C by 1) opening valves **4406** on either side of adjacent pens **4404a** and **4404b**, 2) pumping horizontal flow channel **4402a** to displace cells C and G, and then 3) pumping vertical flow channel **4402b** to remove cell C. FIG. 18D shows that second cell G is moved back into its prior position in cell pen array **4400** by reversing the direction of liquid flow through horizontal flow channel **4402a**. The cell pen array **4404** described

above is capable of storing materials within a selected, addressable position for ready access.

While the embodiment shown and described above in connection with FIGS. 18A-18D utilizes linked valve pairs on opposite sides of the flow channel intersections, this is not required by the present invention. Other configurations, including linking of adjacent valves of an intersection, or independent actuation of each valve surrounding an intersection, are possible to provide the desired flow characteristics. With the independent valve actuation approach however, it should be recognized that separate control structures would be utilized for each valve, complicating device layout.

10. Cell Cage

The cell pen array **4404** described above is capable of storing materials within a selected, addressable position for ready access. However, living organisms such as cells may require a continuous intake of foods and expulsion of wastes in order to remain viable. Accordingly, FIGS. 19A and 19B show plan and cross-sectional views (along line **45B-45B'**) respectively, of one embodiment of a cell cage structure in accordance with the present invention.

Cell cage **4500** is formed as an enlarged portion **4500a** of a flow channel **4501** in an elastomeric block **4503** in contact with substrate **4505**. Cell cage **4500** is similar to an individual cell pen as described above in FIGS. 18A-18D, except that ends **4500b** and **4500c** of cell cage **4500** do not completely enclose interior region **4500a**. Rather, ends **4500a** and **4500b** of cage **4500** are formed by a plurality of retractable pillars **4502**.

Specifically, control channel **4504** overlies pillars **4502**. When the pressure in control channel **4504** is reduced, elastomeric pillars **4502** are drawn upward into control channel **4504**, thereby opening end **4500b** of cell cage **4500** and permitting a cell to enter. Upon elevation of pressure in control channel **4504**, pillars **4502** relax downward against substrate **4505** and prevent a cell from exiting cage **4500**.

Elastomeric pillars **4502** are of a sufficient size and number to prevent movement of a cell out of cage **4500**, but also include gaps **4508** which allow the flow of nutrients into cage interior **4500a** in order to sustain cell(s) stored therein. Pillars **4502** on opposite end **4500c** are similarly configured beneath second control channel **4506** to permit opening of the cage and removal of the cell as desired.

11. Cross-Channel Injector

The cross-flow channel architecture illustrated shown in FIGS. 18A-18D can be used to perform functions other than the cell pen just described. For example, the cross-flow channel architecture can be utilized in mixing applications.

This is shown in FIGS. 20A-D, which illustrate a plan view of mixing steps performed by a microfabricated structures in accordance another embodiment of the present invention. Specifically, portion **7400** of a microfabricated mixing structure comprises first flow channel **7402** orthogonal to and intersecting with second flow channel **7404**. Control channels **7406** overlie flow channels **7402** and **7404** and form valve pairs **7408a-b** and **7408c-d** that surround each intersection **7412**.

As shown in FIG. 20A, valve pair **7408c-d** is initially opened while valve pair **7408a-b** is closed, and fluid sample **7410** is flowed to intersection **7412** through flow channel **7404**. Valve pair **7408a-b** is then actuated, trapping fluid sample **7410** at intersection **7412**.

Next, as shown in FIG. 20B, valve pairs **7408c-d** are closed and **7408a-b** are opened, such that fluid sample **7410** is injected from intersection **7412** into flow channel **7402** bearing a cross-flow of fluid. The process shown in FIGS. 20A-B

can be repeated to accurately dispense any number of fluid samples down cross-flow channel **7402**.

While the embodiment of a process-channel flow injector structure shown in FIGS. **20A-B** feature channels intersecting at a single junction, this is not required by the present invention. Thus FIG. **20C** shows a simplified plan view of another embodiment of an injection structure in accordance with the present invention, wherein junction **7450** between intersecting flow channels **7452** is extended to provide additional volume capacity. FIG. **20D** shows a simplified plan view of yet another embodiment of an injection structure in accordance with the present invention, wherein elongated junction **7460** between intersecting flow channels **7462** includes branches **7464** to provide still more injection volume capacity.

12. Rotary Mixing Structure

Microfluidic control and flow channels in accordance with embodiments of the present invention may be oriented to rotary pump design which circulates fluid through a closed circuit flow channel. As used herein the term “closed circuit” has the meaning known in the art and refers to configurations that are circular and variations thereof such as ellipsoids and ovals, as well as flow circuit paths having corners as are created by triangular, rectangular, or more complex shapes.

As illustrated in FIG. **21**, a layer with flow channels **2100** has a plurality of sample inputs **2102**, a mixing T-junction **2104**, a central circulation loop **2106** (i.e., the substantially circular flow channel), and an output channel **2108**. The overlay of control channels with a flow channel can form a microvalve. This is so because the control and flow channels are separated by a thin elastomeric membrane that can be deflected into the flow channel or retracted therefrom.

The substantially circular central loop and the control channels that intersect with it form the central part of the rotary pump. The pump(s) which cause solution to be flowed through the substantially circular flow channel consist of a set of at least three control channels **2110a-c** that are adjacent to one another and which intersect the substantially circular branch flow channel **2106** (i.e., the central loop).

When a series of on/off actuation sequences, such as 001, 011, 010, 110, 100, 101, are applied to the control channels, the fluid in the central loop can be peristaltically pumped in a chosen direction, either clockwise or counterclockwise. The peristaltic pumping action results from the sequential deflection of the membranes separating the control channels and flow channel into or out of the flow channel.

In general, the higher the actuation frequency, the faster the fluid rotates through the central loop. However, a point of saturation may eventually be reached at which increased frequency does not result in faster fluid flow. This is primarily due to limitations in the rate at which the membrane can return to an unactuated position.

While the system shown in FIG. **21** shows each pump including three control channels, a different number of control channels can be utilized, for example, a single serpentine control channel having multiple cross-over points could be used.

A variety of different auxiliary flow channels which are in fluid communication with the central loop can be utilized to introduce and withdrawn sample and reactant solutions from the central loop. Similarly, one or more exit or outlet flow channels in fluid communication with the central loop can be utilized to remove solution from the central loop. For example, control valves can be utilized at the inlet(s) and the outlet(s) to prevent solution flow into or out from the central loop.

Flow channel sizes and shapes can vary. With certain devices, the diameter of the channel tends to range from about 1 mm to 2 cm, although the diameter can be considerably larger in certain devices (e.g., 4, 6, 8, or 10 cm). Limits on how small the diameter of the circular flow channel can be are primarily a function of the limits imposed by the multilayer soft lithography processes. Channel widths (either flow or control) usually vary between 30 μm and 250 μm. However, channel width in some devices is as narrow as 1 μm. Channels of larger widths can also be utilized, but generally require some type of structural support within the flow channel. Channel height generally varies between 5 and 50 μm. In flow channels having a width of 100 μm or less, the channel height may be 1 μm or smaller. The flow channel is typically rounded to allow for complete blockage of the channel once the membrane is deflected into the channel. In some devices, the channels have shapes such as octagons or hexagons. In certain devices, the flow channels are rounded and 100 μm wide and 10 μm high and control channels are 100 μm wide and 10 μm high. One system that has been utilized in certain studies has utilized a central loop having a diameter of 2 cm, a flow channel width of 100 μm and a depth of 10 μm.

While the channels typically have the foregoing sizes and shapes, it should be recognized that the devices provided herein are not limited to these particular sizes and shapes. For example, branches present in a closed circuit flow channel may serve to control the dispersion and hence mixing of materials flowed therein.

13. Microfluidic Large-Scale Integration

The previous section has described monolithic microvalves that are substantially leakproof and scalable, and has also described methods for fabricating these microvalves. For the relatively simple assemblies of microfluidic valves previously described, each fluid flow channel may be controlled by its own individual valve control channel. However, such a non-integrated control strategy cannot be practicably implemented for more complex assemblies comprising thousands or even tens of thousands of individually addressable valves. Accordingly, a variety of techniques may be applied alone or in combination to allow for the fabrication of large scale integrated microfluidic devices having individually addressable valves.

Techniques useful for implementing large scale integrated microfluidic structures in accordance with embodiments of the present invention are discussed in detail in pending U.S. nonprovisional patent application Ser. No. 10/670,997. One technique allowing for the fabrication of large scale integrated microfluidic devices is the use of multiplexor structures.

The use of multiplexor structures has previously been described in connection with a single set of control lines overlying a single set of flow channels. FIG. **22A** shows a simplified plan view illustrating a microfluidic binary tree multiplexor operational diagram. Flow channels **1900** defined in a lower elastomer layer contain the fluid of interest, while control channels **1902** defined in an overlying elastomer layer represent control lines containing an actuation fluid such as air or water. Valves **1904** are defined by the membranes formed at the intersection of the wider portion **1902a** of a control channel **1902** with a flow channel **1900**. The actuation pressure is chosen so that only the wide membranes are fully deflected into the flow channel **1900**. Specifically, the multiplexor structure is based on the sharp increase in pressure required to actuate a valve as the ratio of control channel width:flow channel width is decreased.

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The multiplexor structure shown in FIG. 22A is in the form of a binary tree of valves where each stage selects one out of two total groups of flow channels. In the multiplexor embodiment shown in FIG. 22A, each combination of open/closed valves in the multiplexor selects for a single channel, so that n flow channels can be addressed with only $2 \log_2 n$ control channels.

By using multiplexed valve systems, the power of the binary system becomes evident: only about 20 control channels are required to specifically address 1024 flow channels. This allows a large number of elastomeric microvalves to perform complex fluidic manipulations within these devices, while the interface between the device and the external environment is simple and robust.

FIG. 22B shows a simplified plan view of an alternative embodiment of a multiplexor structure in accordance with the present invention. Multiplexor structure 1950 comprises control channels 1952 formed in an elastomer layer overlying flow channels 1954 of an underlying elastomer layer. Operating under the same physical principles of the multiplexor of FIG. 22A, multiplexor 1950 comprises a tertiary tree of valves, where each stage comprises three bits ("a trit") and selects one out of three total groups of flow channels. Each combination of open/closed valves in the multiplexor 1950 selects for a single channel, so that n flow channels can be addressed with only $3 \log_3 n$ control channels.

The general microfluidic flow architecture of either of the basic multiplexor devices shown in FIGS. 22A-B may be generically represented in the simplified cross-sectional view of FIG. 22C, wherein second elastomer layer E2 defining control channel network C overlies first elastomer layer E1 defining flow channel network F.

The base 3 multiplexor of FIG. 22B is the most efficient design that may be used to address large numbers of "flow" channels. This is because the $x \log_e n$ valve is minimized where e is used for the base of the log. As fractions are not used for the base of an actual multiplexor, the most efficient multiplexor structure is achieved where the value of $x=3$, the integer closest to e (~2.71828).

To highlight this point, TABLE 2 compares the efficiency of the base 2 multiplexor with the base 3 multiplexor.

TABLE 2

| Number of Control Lines | Number of Flow Lines Controlled by Control Lines | | Enhanced Efficiency of Base 3 Multiplexor Structure |
|-------------------------|--|--------------------|---|
| | Base 2 Multiplexor | Base 3 Multiplexor | |
| 6 | 8 | 9 | +1 |
| 9 | 23 | 27 | +4 |
| 12 | 64 | 81 | +17 |
| 15 | 181 | 243 | +62 |
| 18 | 512 | 729 | +217 |

While the above description has focused upon various multiplexor structures utilizing stages having the same base number, this is not required by the present invention. Alternative embodiments of multiplexor structures in accordance with the present invention may comprise stages of unlike base numbers. For example, a two-stage multiplexor consisting of a bit stage and a trit stage represents the most efficient way of addressing six flow channels. The order of the stages is arbitrary, and will always result in the same number of flow lines being controlled. The use of multiplexor structures comprising different binary and tertiary stages allows the efficient addressing of any number of "flow" channels that are the product of the numbers 2 and 3.

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A multiplexor may conceivably use any base number. For example, five may also be used as the base number, if necessary. However, efficiency in utilization of control lines diminishes as the number of control lines moves away from the value of e . This is shown in FIG. 23, which plots the number of control lines versus the number of flow lines being controlled, for multiplexor structures having different base numbers.

Another technique allowing for the fabrication of large scale integrated (LSI) microfluidic devices is the use of multiple layers of control lines. FIGS. 24-24B illustrate this approach. FIG. 24 shows a plan view of one embodiment of a microfluidic device having a first control line controlled by a second control line. FIG. 24A shows a cross-sectional view of the microfluidic device of FIG. 24, taken along line A-A'. FIG. 24B shows a cross-sectional view of the microfluidic device of FIG. 24, taken along line B-B'.

Microfluidic structure 2100 comprises two flow channels 2102a-b formed in lowermost elastomer layer 2104. First control channel network 2106 including first inlet 2106a in fluid communication with first and second branches 2106b and 2106c, is formed in a second elastomer layer 2108 overlying first elastomer layer 2104. First branch 2106b of first control channel network 2106 includes widened portion 2110 overlying first flow channel 2102a to define first valve 2112. Second branch 2106c of first control channel network 2106 includes widened portion 2114 overlying second flow channel 2102b to define second valve 2116.

Second control channel network 2118 comprising third control channel 2118a is formed in third elastomer layer 2120 overlying second elastomer layer 2108. Third control channel 2118a includes widened portion 2118b overlying first branch 2106b of first control channel network 2106 to form valve 2122.

The microfluidic device illustrated in FIGS. 24-24B may be operated as follows. A fluid that is to be manipulated is present in flow channels 2102a and 2102b. Application of a pressure to the first control channel network 2106 causes the membranes of valves 2112 and 2116 to deflect downward into their respective flow channels 2102a and 2102b, thereby valving flow through the flow channels.

Application of a pressure to second control channel network 2118 causes the membrane of valve 2122 to deflect downward into underlying first branch 2106c only of first control channel network 2106. This fixes the valve 2112 in its deflected state, in turn allowing the pressure within the first control channel network 2106 to be varied without affecting the state of valve 2112.

The general architecture of the microfluidic device depicted in FIGS. 24-24B is summarized in the simplified cross-sectional view of FIG. 25. Specifically, elastomeric device 2200 comprises lowest elastomer layer E1 defining flow channel network F, underlying second elastomer layer E2 defining first control channel network C1. First control channel network C1 in turn underlies second control channel network C2 that is defined within third elastomer layer E3.

While the embodiment of the microfluidic device of FIGS. 24-24B is described as being fabricated from three separate elastomer layers, this is not required by the present invention. Large scale integrated microfluidic structures in accordance with embodiments of the present invention featuring multiplexed control lines may be fabricated utilizing only two elastomer layers. This approach is shown and illustrated in connection with FIGS. 26-26B.

FIG. 26 shows a simplified plan view of a microfabricated elastomer device including first and second flow channels 2300a and 2300b, and first branched control channel network

2302 overlying flow channels 2300a and 2300b to define valves 2304 and 2306 respectively. FIG. 26A shows a cross-sectional view of the microfabricated elastomer device of FIG. 26, taken along line A-A', with flow channel 2300a defined in lower elastomer layer 2308, and first control channel 2302 defined in upper elastomer layer 2310.

Lower elastomer layer 2308 further comprises a second control channel network 2312 running underneath first control channel 2302 to define valve 2314. Accordingly, FIG. 26B shows a cross-sectional view of the microfabricated elastomer device of FIG. 26, taken along line B-B'. While present in the same (lower) elastomer layer 2308, flow channel network 2300 and second control channel network 2312 are separate and do not intersect one another.

As represented in the simplified cross-sectional view of FIG. 27, separate flow channel network F and control channel network C2 may thus be present on a single (lower) elastomer layer E1 that is overlaid by another elastomer layer E2 defining only a control channel network C1.

The microfluidic device illustrated in FIGS. 26-26B may be operated as follows. A fluid that is to be manipulated is present in flow channels 2300a and 2300b. Application of a pressure to the first control channel network 2302 causes the membranes of valves 2304 to deflect downward into their respective flow channels 2300a and 2300b, thereby valving flow through the flow channels.

Application of a pressure to second control channel network 2312 causes the membrane of valve 2314 to deflect upward into the overlying branch of first control channel network 2302. This fixes the valve 2314 in its deflected state, in turn allowing the pressure within the first control network 2302 to be varied without affecting the state of valve 2314. In contrast with the embodiment shown in FIG. 24, the microfluidic device of FIGS. 26-26B features a valve that operates by deflecting upward into an adjacent control channel in response to an elevated pressure.

FIG. 27A shows a simplified schematic plan view of a microfluidic comparator chip 3000 microfabricated with large scale integration technology which is analogous to an array of 256 comparators. Specifically, a second device containing 2056 microvalves was designed which is capable of performing more complex fluidic manipulations. The various inputs have been loaded with colored food dyes to visualize the channels and sub-elements of the fluidic logic.

Comparator chip 3000 is formed from a pair of parallel, serpentine flow channels 3002 and 3004 having inlets 3002a and 3004a respectively, and having outlets 3002b and 3004b respectively, that are intersected at various points by branched horizontal rows of flow channels 3006. Portions of the horizontal flow channels located between the serpentine flow channels define mixing locations 3010.

A first barrier control line 3012 overlying the center of the connecting channels is actuable to create adjacent chambers, and is deactivable to allow the contents of the adjacent chambers to mix. A second barrier control line 3014 doubles back over either end of the adjacent chambers to isolate them from the rest of the horizontal flow channels.

One end 3006a of the connecting horizontal flow channel 3006 is in fluid communication with pressure source 3016, and the other end 3006b of the connecting horizontal flow channel 3006 is in fluid communication with a sample collection output 3018 through multiplexor 3020.

FIGS. 27B-G show simplified enlarged plan views of operation of one mixing element of the structure of FIG. 27A. FIG. 27B shows the mixing element prior to loading, with the mixer barrier control line and wrap-around barrier control line unpressurized. FIG. 27C shows pressurization of the

wrap-around barrier control line and barrier mixer line to activate isolation valves and separation valve to define adjacent chambers 3050 and 3052. FIG. 27D shows loading of the chambers with a first component and a second component by flowing these materials down the respective flow channels. FIG. 27E shows pressurization of the vertical compartmentalization control line 3025 and the isolation to define the adjacent chambers.

FIG. 27F shows depressurization of the mixing barrier control channel to deactivate the separation barrier valve, thereby allowing the different components present in the adjacent chambers to mix freely.

FIG. 27G shows the deactivation of barrier the isolation control line, causing deactivation of the isolation valves, followed by application of pressure to the control line and deactivation of the multiplexor to allow the combined mixture to be recovered.

In the case of the device shown in FIGS. 27A-G, two different reagents can be separately loaded, mixed pair wise, and selectively recovered, making it possible to perform distinct assays in 256 sub-nanoliter reaction chambers and then recover a particularly interesting reagent. The microchannel layout consists of four central columns in the flow layer consisting of 64 chambers per column, with each chamber containing ~750 pL of liquid after compartmentalization and mixing. Liquid is loaded into these columns through two separate inputs under low external pressure (~20 kPa), filling up the array in a serpentine fashion. Barrier valves on the control layer function to isolate the sample fluids from each other and from channel networks on the flow layer used to recover the contents of each individual chamber. These networks function under the control of a multiplexor and several other control valves.

The storage array and comparator microfluidic devices shown in FIGS. 27A-G was fabricated with multilayer soft lithography techniques using two distinct layers. The "control" layer, which harbors all channels required to actuate the valves, is situated on top of the "flow" layer, which contains the network of channels being controlled. A valve is created where a control channel crosses a flow channel. The resulting thin membrane in the junction between the two channels can be deflected by hydraulic or pneumatic actuation. All biological assays and fluid manipulations are performed on the "flow" layer.

Master molds for the microfluidic channels were made by spin-coating positive photoresist (Shipley SJR 5740) on silicon 9 μm high and patterning them with high resolution (3386 dpi) transparency masks. The channels on the photoresist molds were rounded at 120° C. for 20 minutes to create a geometry that allows full valve closure.

The devices were fabricated by bonding together two layers of two-part cure silicone (Dow Corning Sylgard 184) cast from the photoresist molds. The bottom layer of the device, containing the "flow" channels, is spin-coated with 20:1 part A:B Sylgard at 2500 rpm for 1 minute. The resulting silicone layer is ~30 μm thick. The top layer of the device, containing the "control" channels, is cast as a thick layer (~0.5 cm thick) using 5:1 part A:B Sylgard using a separate mold. The two layers are initially cured for 30 minutes at 80° C.

Control channel interconnect holes are then punched through the thick layer (released from the mold), after which it is sealed, channel side down, on the thin layer, aligning the respective channel networks. Bonding between the assembled layers is accomplished by curing the devices for an additional 45-60 minutes at 80° C. The resulting multilayer devices are cut to size and mounted on RCA cleaned No. 1, 25 mm square glass coverslips, or onto coverslips spin coated

with 5:1 part A:B Sylgard at 5000 rpm and cured at 80° C. for 30 minutes, followed by incubation at 80° C. overnight.

Simultaneous addressing of multiple non-contiguous flow channels is accomplished by fabricating control channels of varying width while keeping the dimension of the flow channel fixed (100 μm wide and 9 μm high). The pneumatic pressure in the control channels required to close the flow channels scales with the width of the control channel, making it simple to actuate 100 μm×100 μm valves at relatively low pressures (~40 kPa) without closing off the 50 μm×100 μm crossover regions, which have a higher actuation threshold.

Introduction of fluid into these devices is accomplished through steel pins inserted into holes punched through the silicone. Unlike micromachined devices made out of hard materials with a high Young's modulus, silicone is soft and forms a tight seal around the input pins, readily accepting pressures of up to 300 kPa without leakage. Computer-controlled external solenoid valves allow actuation of multiplexors, which in turn allow complex addressing of a large number of microvalves.

II. Microfluidic Matrix Architecture

FIG. 28A shows a simplified schematic plan view of one embodiment of a microfluidic matrix architecture 6000 in accordance with the present invention, showing the N=20 matrix chip layout, and the various input, output and control ports. The scale bar of FIG. 28A is 6.4 mm. FIG. 28B shows an enlarged plan view of one reactor from the upper left hand corner of the matrix shown in FIG. 28A.

The apparatus of FIG. 28A features alternating row flow channels 6048 and 6049, intersecting with alternating column flow channels 6050 and 6052, with alternative pairs of successive flow channels 6050 and 6052 linked by branch flow channels 6086 to define reactor sites at the flow channel vertices. Each of flow channel row 6048 includes a common inlet port 6048a and an outlet port 6048b. Each of flow channel rows 6049 includes a common inlet port 6049a and an outlet port 6049b. As described below, in certain applications relating to the amplification of nucleic acids, flow channel rows 6048 may contain a primer, while flow channel rows 6049 may contain a no-primer control.

Each of alternate flow channel columns 6050 include an inlet port 6050a and an outlet port 6050b. Alternate flow channel columns 6052 include a common inlet port 6052a and an outlet port 6052b.

Each reactor 6002 of matrix 6000 is in pressure communication with control lines present in underlying and overlying elastomer layers. Specifically, First control lines 6016 overlap column flow channels 6050 and 6052, thereby defining valves 6016a. First control lines 6016 are in fluid communication with common port 6016a. Second control lines 6035 overlap row flow channels 6048 and branch flow channels 6086, thereby defining valves 6036 and 6038 having different widths. Second control lines 6035 are in fluid communication with common port 6035a.

First pump line 6018 underlies column flow channel 6052, thereby defining first push-up pumps 6019. Various separate groupings of first pump line 6018 are in fluid communication with common ports 6018a. Second pump line 6022 underlies branch flow channel 6086, thereby defining second push-up pumps 6024. Various groupings of second pump line 6022 are in fluid communication with common ports 6022a.

FIG. 28B shows an enlarged plan view of a single reactor 6002 featuring relatively large control lines 6016 including stabilizing post structures 6088, column flow line 6050 containing a first template sample 6008, a column flow line 6052 containing DNA polymerase 6010, and a row flow line 6049 containing primer 6012 or a no-primer control. The respective

volume ratios are $V_{template}:V_{polymerase}:V_{primers}=3:3:4$. Rotary pumping valves 6019 and 6024 are defined between reactor 6006 and underlying pumping control channels 6018 and 6022.

FIG. 28C shows a simplified cross-sectional view of the microfluidic matrix of FIG. 28B taken along line 28C-28C'. FIG. 28D shows a simplified cross-sectional view of the microfluidic matrix of FIG. 28D taken along line 28D-28D'.

As shown in FIGS. 28C-D, the device 6000 comprises three layers 6060, 6062, and 6064 with distinct functions. Layer 6062 containing matrix flow channel structure 6006 is sandwiched between two layers 6060 and 6064 containing integrated hydraulic valves 6016a and pneumatic pumps 6019 and 6024, respectively.

In the middle fluidic layer 6062, flow channels 6006 are 106 pm wide and 12~14 μm high. Each vertex of the matrix 6000 contains a ring shaped channel (reactor) 6002 of approximate 3 nanoliters. Each row of reactors 6002 is connected to a separate input hole 6048a or 6049a (about 625 μm in diameter) through which unique primers or other reagents may be loaded. Each column can similarly load the reactors with different DNA templates or other reagents through second input holes 6050a. A third, single common input 6052a for the addition of polymerase or another reagent is connected to all the reactors in the matrix.

In the top, control layer 6060, the valve system is designed to load each reactor with the three separate reagents while preventing cross contamination. In total, 2860 valves displayed horizontally or vertically are controlled by only two independent pressure supply through-holes 6016b and 6035b. Furthermore, the large valves 6038 (270 μm wide) or the small valves 6036 (96 μm wide) can selectively be actuated because they have a different threshold of the hydraulic pressure. This is described by Unger et al. in *Science*, 288, 113-116 (2000) and Thorsen et al. in *Science*, 298, 580-584 (2002). As described by Unger et al., reagent loading is not blocked by the narrow control channels 6035a (42 μm wide) connecting the valve system, because their tiny membrane does not deflect at the actuation pressure used.

The third, bottom layer 6064 uses a 20×20 array of rotary pumps in order to facilitate mixing the reagents. The use of such devices is described by Chou et al., *Biomedical Microdevices*, 3, 323 (2001), and by Liu et al., *Electrophoresis*, 23, 1531-1536 (2002). In the specific embodiment illustrated in FIGS. 28A-D, each reactor 6002 features only two rotary pump structures 6019 and 6024. This represents a simplification over alternative embodiments of microfluidic matrix structures in accordance with the present invention, wherein three such pumping structures underlie each rotary flow channel reactor. A minimum of only two such pumping structures are required to achieve peristaltic pumping, owing to the closed-loop nature of the rotary flow channel reactor at the vertices.

FIGS. 28E-H show enlarged and simplified plan views of a group of four adjacent reactors in the matrix of FIGS. 28A-D, illustrating successive steps of operation of the microfluidic matrix device.

As shown in FIGS. 28E-H, active valves in the top control structure 6060 facilitate the loading and isolation of reagents 6008, 6010, and 6012. Specifically, as shown in FIG. 28E, with valves 6016a actuated, primer sets or other reagents 6012 are loaded along rows 6050 and 6052 respectively, of the matrix. Actuation of valves 6036 and 6038 isolates a well-defined volume of primers or other reagents 6012 in each reactor 6002.

As shown in FIG. 28F, valves 6036 and 6038 are closed and valves 6016a opened, to allow for the loading of DNA tem-

plates or other reagents **6008** down alternate columns **6050** while polymerase or another reagent **6010** is simultaneously introduced to all reactors from a single inlet **6052a** through alternate columns **6052**. Valves **6016a** are then once again actuated, defining the desired volumes of the different reagents **6008** and **6010** such as polymerase and templates, and isolating each reactor **6006a**.

As described by Unger et al., the different membrane areas of valves **6036** and **6038** allow for the selective opening of only valves **6036** by reducing the actuation pressure from 260 kPa to 110 kPa, thus bringing all three components **6008**, **6010**, and **6012** into fluidic contact within each reactor **6002**. As shown in FIG. **28G**, peristaltic pumps **6019** and **6024** in the bottom layer allow for the rapid rotary mixing of all reagents within the reactors **6002**.

The embodiment of FIGS. **28A-H** features pumping control lines organized in groups sharing different common inlets/outlets, thereby reducing accumulated membrane delay.

The protocols of fabricating two-layer polydimethylsiloxane (PDMS) devices using multilayer soft lithography (MSL) have been previously described, at least by Fu et al., and Hansen et al. For the three-layer microfluidic matrix elastomeric devices in accordance with embodiments of the present invention, three separate molds for the different layers were prepared with UV-lithography. Photoresist of each mold layer may be heated to promote reflow and corresponding formation of rounded features.

The photoresist (Shipley SJR5740) line height was 12-44 μm on the fluidic layer mold or the pump layer mold. A 25 μm thick elastomeric layer with a 20:1 ratio was spin-coated on the fluidic layer mold at 1600 RPM for 65 seconds, then cured at 80° C. for 35 minutes.

The mold for the control layer had higher lines of 25 μm , which prevents collapse of the thin elastomeric membrane of the fluidic layer. To fabricate the control layer, General Electric RTV 615 A and B components were mixed in a 4:1 ratio using a Keyence Hybrid mixer. Then the mixture was poured onto the control layer mold in an foil-coated petri dish and baked in the oven at 80° C. for 30 minutes to obtain a thick PDMS block (~3-4 mm).

After incubation, the control layer was peeled from the mold and punched with through-holes for pressure connection. It was aligned on the thin fluidic layer and then baked at 80° C. for 45 minutes.

A third pump layer was spin-coated with the RTV615 mixture (20:1) at 2200 RPM for 65 seconds, and then incubated at 80° C. for 45 minutes. Next, the bonded device containing the former two control and fluidic layers was peeled and through-holes were punched for loading fluidic samples.

The bonded device was then aligned onto the third pump layer and baked at 80° C. overnight. Then the rest through-holes were punched on the device for air injection in pumping. After that, the three-layer device was sealed with a piece of glass coverslip (#1, Lakeside microscope accessories) and incubated at 80° C. for no less than 3 hours.

III. PCR

Nucleic acid amplification reactions have emerged as powerful tools in a variety of genetic analyses and diagnostic applications. The value of these techniques is their ability to rapidly increase the concentration of target nucleic acids of interest that might be present at very low and otherwise undetectable levels. For instance, by utilizing the polymerase chain reaction (PCR) amplification technique, one can amplify a single molecule of a target nucleic acid by 10^6 to 10^9 .

PCR is perhaps the most well-known of a number of different amplification techniques. This well established procedure involves the repetition of heating (denaturation) and cooling (annealing) cycles in the presence of a target nucleic acid, primers that hybridize to the target, deoxynucleotides, a polymerase and cofactors such as metal ions. Each cycle produces a doubling of the amount of the target DNA. The cycles are conducted at characteristic temperatures: 95° C. for denaturing double stranded nucleic acid, 50 to 65° C. for hybridization of primer to the target nucleic acid, and 72 to 77° C. for primer extension (see, generally, Sambrook et al., *Molecular Cloning: A Laboratory Manual*, 2nd ed., Cold Spring Harbor Laboratory Press (1989), incorporated by reference herein for all purposes. See also for example, U.S. Pat. Nos. 4,683,202 and 4,683,195, both of which are incorporated by reference herein for all purposes.

Methods for conducting PCR amplifications fall into two general classes. The approach typically utilized is a time domain approach in which the amplification reaction mixture is kept stationary and the temperature is cycled. See, e.g., Cheng, et al., *Nucleic Acids Res.* 24:380-385 (1996); Shoffer, et al., *Nucleic Acids Res.* 24:375-379 (1996); and Hong, et al., *Electrophoresis* 22:328-333 (2001), each of which is incorporated herein by reference for all purposes. While methods utilizing this approach can be conducted with relatively small sample volumes, the methods require complex regulation of heater elements and relatively long reaction times.

Another approach that has been discussed is limited to a space domain approach in which three temperature zones are constantly kept at the different temperatures and the reaction mixture runs in a serpentine flow channel above it. See, e.g., Kopp et al., *Science* 280:1046-1048 (1998), incorporated by reference herein for all purposes. A method such as this can be conducted at relatively high speed because it is not necessary to heat and cool the heaters, but requires the use of relatively large sample volumes.

FIG. **34** shows a schematic diagram of one embodiment of a microfluidic structure **6300** for performing PCR. Top layer **6302** includes control channels **6304a-d** for controlling the pumps and valves. Middle layer **6304** defines the inlet **6305a**, rotary **6305b**, and outlet **6305c** fluid flow channels. Bottom layer **6306** includes integrated heater structures **6308** and electrical leads **6310** in electrical communication therewith.

The loop in the fluid layer forms a rotary pump, by which the PCR reagents can be transported over regions of different temperatures. The temperatures are set by tungsten heaters evaporated onto glass, which become the bottom substrate of the nanofluidic chip. The total volume of the PCR reaction was 12 nL.

FIG. **35** shows results of Taqman PCR assay performed in the rotary pump chip. Fluorescence was measured in situ at various time points as the PCR reaction mixture was pumped through different temperature regions. A fragment of the β -actin gene was amplified from human male genomic DNA. The closed circles represent data from an experiment which contained the human template DNA. The open circles represent data from a negative control experiment in which template DNA was withheld.

Amplification of purified nucleic acid samples in accordance with embodiments of the present invention is not limited to the specific microfluidic structure shown in FIG. **34**. Alternative, more complex microfluidic geometries are possible.

For example, the microfluidic matrix device shown in FIGS. **28A-H** could be utilized to perform PCR. In order to test the performance of the microfluidic matrix chips just described, a number of experiments were performed.

Specifically, each row of reactors was alternatively loaded with primers and no-primer controls. Each column of reactors was then loaded with the cDNA template and no-template controls. The DNA template concentration was titrated along the positive control columns in order to test sensitivity.

The polymerase was loaded from the single inlet and amortized over all the independent reactors. The reactors containing all the necessary reagent components for PCR produced the positive signals (yellow, a mixed color of green and dim red); while those reactors that missed one or two reagent components showed a color of dim red, which derived from the passive reference dye in the buffer solution. The expected image is a checkerboard pattern, and any cross-contamination or leakage between compartments will be evident.

In these experiments, a 294-bp segment of the human (3-Actin cDNA fragment (1.8 Kbp, Clontech) was amplified. Forward primer and reverse primer were the following: 5'-TCA CCC ACA CTG TGC CCA TCT ACG A-3' and 5'-CAG CGG AAC CGC TCA TTG CCA ATG G-3'. An amplification protocol supplied by Applied Biosystems (<http://www.appliedbiosystems.com>) was modified to include the use of DyNAzyme II (0.1 U/reaction, Finnzymes OY) and additives. See Yang et al., *Lab on a Chip*, 2, 179-187 (2002). The solutions loaded onto the chip were prepared with one component of the template, primers or polymerase, and all other necessary PCR reagents and additives.

To simplify the pipetting steps for the checkerboard experiments, two sets of $N/2$ inputs/outputs are connected together for primer loading. The volumes loaded were 3 μ l primers and no-primer control, 1 μ l cDNA or no-template control, and 2 μ l Dynazyme. An efficient mixing of the reagents was achieved by actuating the rotary pumps with two pneumatic controllers (Fluidigm Inc., So. San Francisco) at 10 Hz for 5 minutes.

After mixing was complete, the matrix chip was transferred onto a flat-bed thermocycler (DNA Engine, MJ Research). All the individual reactors were still isolated by the hydraulic control channels, which reduced the evaporation of the reagents in the fluidic channels to a negligible level.

The whole chip was thermocycled in the following protocol: 2 minutes at 50° C., and 1 minute at 96° C.; then 30 cycles: 20 seconds at 96° C.; 40 seconds at 60° C. The PCR conditions were verified with conventional methods to show that there are no detectable side products from the non-hot start condition. After thermocycling, the fluorescence emission from the PCR products in the chip (519 nm and 570 nm) was imaged using a modified DNA array scanner (Applied Precision).

FIG. 29 shows a two-color image of fluorescent emission from a 20 \times 20 matrix chip. Yellow indicates a positive signal from the sample. The no-template or no-primer controls show a dim red produced by the passive reference dye in the buffer. The concentration of the DNA templates is as follows (copies per reactor): C1&C3 (6,100), C5&C7 (3,050), C9&C11 (610), C13&C15 (305), C17&C19 (61). A threshold fluorescent ratio is established to define the false positive/negative signal. Some reactors in columns 17&19 do not show the desired positive signals, establishing the sensitivity of the chip. The scale bar of FIG. 29 is 6.4 mm.

FIG. 30 shows a scanned fluorescent image illustrating use of a 10 \times 10 matrix chip in which an alternative reagent format was used. The scale bar of FIG. 30 is 6.4 mm.

Specifically, the matrix chip of FIG. 30 was used to explore combinations of forward and reverse primers. The three main components with all the other necessary PCR reagents and additives were as follows: component I included both the cDNA template and the polymerase; components II con-

tained forward primers while component III contained reverse primers. Components II and III were loaded in the rows and columns of the matrix, respectively. Then component I, which contained both the cDNA template and the polymerase, was loaded into all the reactors from the single input.

Even-numbered rows are loaded with the correct forward primers, while odd rows are negative controls. Columns 1, 3, 5, 6, 8, and 10 are loaded with the correct reverse primers, while the remainder columns are negative controls. Thus, one should only observe successful PCR reactions in those reactors at the vertices of correct forward and reverse primers, as shown. Reactor C6R9 shows a false positive signal, which possibly resulted from the contamination of the forward primer of the neighboring reactors in the same column (C6).

In this manner, all possible combinations of forward and reverse primers can be tested with each other on a common DNA template. This format is particularly useful for PCR optimization and exon mapping.

The chip layout and sample loading sequence are specifically designed so that DNA template contamination among independent reactors is only possible if the valve system fails. Although there exists in principle a chance of slight primer contamination, the absolute numbers are so low that such an effect should not be expected to affect the outcome of the assay. Possible primer contamination can be further reduced by flushing buffer solution into the fluidic channels before loading the DNA templates or polymerase.

In all, six separate 20 \times 20 matrix chips and 8 separate 10 \times 10 matrix chips were tested for a total of 3,200 reactions. Twenty-four out of 1550 positive control reactors produced false negative signals. In eighteen of these cases, the DNA concentrations were close to the detection limit. The other six reactors failed, possibly because of dust blocking the channels or other fabrication defects.

Thirty-five out of 1,650 negative controls showed false positive signals, perhaps due to the reagent leakage. In total, 98% of the 3,200 reactors tested produced the expected results.

By titrating the template DNA concentration, the detection limit of the chip was established to be around 60 template copies per reactor. FIG. 31 plots fluorescent ratio vs. the initial concentration of the cDNA templates. Each symbol represents the averaged ratio of 20 reactors containing the same template concentration, with an error bar showing the standard deviation. There is no significant difference between the no-template control and the no-primer control. As the concentration decreases to the limit of detection, the standard deviation increases due to a mixture of positive and false negative results.

An immediate application of the microfluidic matrix device exists in medical diagnostics and genetic testing, where the matrix chip allows N patients to be screened for N mutations or pathogens with PCR or other techniques such as fluorescent-labeled mismatch-binding protein. See Behrens-dorf et al., *Nucleic Acids Res.*, 30, e64 (2002).

Another application of interest is gene expression analysis, where reverse transcriptase PCR can be used to query N mRNA samples for the expression levels of N different genes.

In another experiment, positive samples were loaded into every reactor in order to collect as much PCR product as possible. The scanned fluorescent image (not shown) demonstrates that all the 400 reactors worked as expected.

For this experiment, the chip pattern was slightly modified to connect together all the inputs/outputs for primer loading/unloading. Thus, the PCR product could be flushed with TE

buffer solution (about 40 μ l) and collected with a pipette tip plugged into the wired outputs.

The solution of the extracted PCR amplicon was concentrated into a volume of about 5 μ l by evaporation. This pooled amplicon was then verified by gel electrophoresis. FIG. 33 shows Agarose gel electrophoresis of the PCR amplicon by the chip or by a microtube. FIG. 33 is displayed in 2% Agarose gel with easy-cast electrophoresis system (Model BIA, VWR), stained with Sybr Green I, and documented by the Kodak electrophoresis system.

The sample of lane 1 of FIG. 32 is 1 μ l PCR amplicon by the microtube. The sample of lane 2 of FIG. 32 is 0.5 μ l PCR amplicon by the microtube. The sample of lane 3 is a negative control by the microtube. The sample of lane 4 is 100 bp DNA ladder. The sample of lane 5 is extracted PCR amplicon (400 reactors \times 3 nl/reactor) by the matrix chip.

Intensity of the pooled amplicon was comparable to the bench-top positive controls (lane 1 and lane 2). In future devices it should be possible to integrate on chip capillary electrophoresis for in situ product analysis beyond the information provided by the Taqman assay.

To summarize, embodiments of microfluidic matrix architectures in accordance with the present invention represent an effective solution to the macroscopic/microfluidic interface issue, and demonstrate how microfluidics can achieve impressive economies of scale in reducing the complexity of pipetting operations. Using an $N\times N$ microfluidic matrix with $N=20$, $N^2=400$ distinct PCR reactions were performed with only $2N+1=41$ pipetting steps, compared with the $3N^2=1,200$ steps required with conventional fluid handling. Each vertex of the matrix has a 3 nanoliter reactor, and a single 2 μ L aliquot of polymerase is amortized over all 400 independent reactions, thus dramatically reducing sample overhead and minimizing reagent consumption.

While the specific embodiment shown and described above performs PCR by the introduction of primers, nucleic acid samples, and enzymes through separate inlets, this is not required by the present invention. In accordance with one alternative embodiment, the three separate inlet channels can be utilized to flow forward primers, reverse primers, and a mixture of nucleic acid sample and enzyme, respectively, into the three inlets to the flow channels of a microfluidic matrix device.

And while the specific embodiment of a microfluidic device shown and described above is fabricated from three layers of elastomer material, this is also not required by the present invention. In accordance with alternative embodiments, control lines of different widths defining membranes of varying size that are actuatable at different applied pressures, could be formed in a single elastomer layer adjacent to the fluidic flow layer. These various control lines could be utilized to perform reactor gating, isolation, and circulation functions.

Moreover, the embodiments of microfluidic devices in accordance with the present invention are not limited to the specific matrix architecture shown and described in connection with FIGS. 28A-H. For example, alternative embodiments of microfluidic devices in accordance with the present invention could include features that allow for recovery of the contents of individual reactors following reaction. FIGS. 35A-D show simplified plan views showing operation of one such alternative embodiment.

FIG. 35A shows an initial step of operation, wherein reactor 6202 of microfluidic matrix device 6200 has been loaded with reagents through column flow channel pairs 6204 and 6206, followed by reaction by circulation within reactor 6202 isolated by actuation of adjacent valves 6208, to form reaction product 6218. The pumping structures responsible for circu-

lating fluid within the reactor 6202 have been omitted from FIGS. 35A-D for clarity of illustration.

The architecture of the microfluidic matrix device shown in FIG. 35A differs from that of prior Figures in two important respects. First, each rotary reactor 6202 is fabricated adjacent to a holding chamber 6210, that is in turn in fluid communication with an additional column flow channel 6212 gated by valves 6214. As described below in connection with FIGS. 35B-C, holding chamber 6210 and recovery flow channel 6212 play a key role in permitting recovery of the contents of individual reactors from the microfluidic matrix.

A second key difference between the architecture of the microfluidic matrix device shown in FIG. 35A and those discussed previously, is that the reactors are formed from pairs of horizontal branch channels connecting alternative pairs of column flow channels, rather than from continuous row flow channels. This architectural aspect becomes important in ensuring efficient and complete transfer of the contents of the reactor to the adjacent holding chamber, as shown in FIG. 35B.

Specifically, leaving valves 6208 open, buffer solution is flowed into row flow channel 6216, thereby displacing reaction product 6218 present in reactor 6002 into holding chamber 6210. Offsetting the position of upper branch flow channel 6220 from row flow channel 6216 in this embodiment, ensures that the contents of the reactor are captured and transferred during this process.

FIG. 35C shows the next step in the process, wherein the reaction product 6218 is isolated in holding chamber 6210, by reactuating valves 6208. FIG. 35D shows the final step in the process, wherein the reaction product 6218 may be flushed from the microfluidic device and recovered at an outlet or other feature, due to the positive pressure within column recovery flow channel 6212.

It should be noted that when valves 6208 or 6214 are opened, some contamination may occur between neighboring reactors. Such contamination may be caused by convective mixing between the contents of adjacent reactors during pumping along the common row flow channel as shown in FIG. 35B. However, the contamination can be reduced to a minimum level by isolating the contents with the neutral buffer solution in the holding chambers.

Contamination may also result from the passive diffusion of materials, for example between adjacent reaction products pumped along a common column flow channel as shown in FIG. 35D. However, the amount of such contamination is slight, and its impact can often be neglected for many applications. For example, a reaction product comprising medium or high molecular weight DNA takes hours to diffuse a distance of hundreds of microns, while the process of recovering the specified product from the chip can be accomplished in only a few seconds.

In some instances, the microfluidic devices described herein can be used as an analytical tool to amplify a target nucleic acid potentially present in a sample and then detect the amplified product to determine whether the target nucleic acid is present or absent in the sample. Thus, amplification serves to enhance the ability to detect target nucleic acids present at low levels. When utilized in this manner, the devices can be used in a wide variety of different applications.

For example, the devices can be used in various diagnostic applications that involve a determination of whether a particular nucleic acid is present in a sample. Hence, samples can be tested for the presence of a particular nucleic acid associated with particular pathogens (e.g., certain viruses, bacteria

or fungi), for instance. The devices can also be utilized for identification purposes, such as in paternity and forensic cases.

The methods and devices provided herein can also be utilized to amplify low levels of nucleic acid for further examination to detect or characterize specific nucleic acids that are correlated with infectious diseases, genetic disorders or cellular disorders (e.g., oncogenes associated with cancer). Genetic disorders are those that involve a specific deletion and/or mutation in genomic DNA. Examples of genetic diseases that can be detected include, but are not limited to, α - and β -thalassemia, cystic fibrosis and sickle cell anemia. Because the devices and methods disclosed herein can utilize very small sample volumes, they are useful in amplifying DNA samples obtained in conjunction with the prenatal diagnosis of genetic disease.

However, the amplification reactions can also be utilized as just one step of a more extensive process involving the diagnostic testing for particular target nucleic acids and in preparing sufficient nucleic acid for use in various genetic engineering applications. Hence, amplified sample can be used in a number of post amplification manipulations and analyses. Examples of such post amplification processes and analyses include, but are not limited to, sequencing of amplified products, cell-typing, DNA fingerprinting and mapping of DNA sequences.

Amplified products can also be generated for use in various genetic engineering applications. For instance, amplified product can be utilized to conduct recombination studies. In other applications, the devices are used to produce target DNA for insertion into a variety of vectors. Such vectors can then be used to transform cells for the production of desired products such as proteins, or nucleic acids in various therapeutic or biotechnological processes.

The microfluidic devices in accordance with embodiments of the present invention can also be utilized to conduct sequencing reactions such as chain termination methods using dideoxynucleotides. Sequencing reactions utilizing the devices disclosed herein can be conducted in different formats. One approach is to conduct four separate sequencing reactions, a separate reaction being conducted in four different thermal cycling devices as provided herein. Each of the four reactions contains target nucleic acid, a primer complementary to the target, a mixture of one dideoxynucleotide (ddNTP) (optionally labeled) with its counterpart deoxynucleotide (dNTP), and the other three dNTPs. Thus, each one of the reactions is conducted with a different ddNTP/dNTP mix. Following completion of the primer extension reactions, the different sized extension products can be separated by capillary gel electrophoresis. This separation can be performed in a separation module as described supra that is integrated with the present devices or in a stand alone capillary gel electrophoresis apparatus.

In addition, embodiments of microfluidic matrix devices can also be utilized in more streamlined formats in which reactions are conducted simultaneously in a single thermal cycling device using differentially labeled dideoxynucleotides. The resulting mixture of chain-terminated reaction products are then separated on a single capillary gel electrophoresis column. The identity of the dideoxynucleotide incorporated into the primer can be determined on the basis of the label.

Restriction digests of nucleic acids can also be conducted utilizing the microfluidic matrix devices in accordance with embodiments of the present invention. Temperature control in such reactions initially involves controlling the temperature within the device at a temperature that promotes the activity

of the restriction enzyme (e.g., 1-3 hours at 30-50° C. depending upon the particular enzyme. The other temperature is selected to promote enzyme deactivation (e.g., 60° C. for 20 minutes).

Analyses to determine the identity of a nucleotide present at a polymorphic site, i.e. the site of variation between allelic sequences such as a single nucleotide polymorphism (SNP), can also be conducted with certain of the present devices. Often these analyses are conducted using single base pair extension (SBPE) reaction. A number of SPBE assays have been developed, but the general approach is quite similar. Typically, these assays involve hybridizing a primer that is complementary to a target nucleic acid such that the 3' end of the primer is immediately 5' of the variant site or is adjacent thereto. Extension is conducted in the presence of one or more labeled non-extendible nucleotides that are complementary to the nucleotide(s) that occupy the variant site and a polymerase. The non-extendible nucleotide is a nucleotide analog that prevents further extension by the polymerase once incorporated into the primer. If the added non-extendible nucleotide(s) is(are) complementary to the nucleotide at the variant site, then a labeled non-extendible nucleotide is incorporated onto the 3' end of the primer to generate a labeled extension product. Hence, extended primers provide an indication of which nucleotide is present at the variant site of a target nucleic acid. Such methods and related methods are discussed, for example, in U.S. Pat. Nos. 5,846,710; 6,004,744; 5,888,819; 5,856,092; and 5,710,028; and in WO 92/16657, each of which is incorporated herein by reference for all purposes.

Using devices as described herein, the temperature within a temperature control region can be selected to promote the primer annealing, primer extension and denaturation steps involved in these particular analyses, and thus allows these extension reactions to be conducted in a thermal cycling format.

IV. Other Applications

Beyond PCR, the matrix chip provides a general method to perform chemical and biological experiments with precious reagents in a highly automated fashion. For example, while the foregoing discussion of the microfluidic matrix devices has focused on their utility in conducting a large number of nucleic acid amplification reactions, it will be appreciated by those with ordinary skill in the art that such microfluidic devices can be utilized to conduct a wide variety of types of reactions and screening methods. Thus, by way of illustration but not limitation, the devices can be utilized to conduct synthetic reactions between a plurality of reactants. Using the device shown in FIGS. 28A-H, for instance, a first set of reagents can be introduced into the horizontal flow channels, a second set of reagents can be introduced into the vertical flow channels that have an independent inlet; and a third reagent can be introduced into the vertical flow channels that are connected to the shared inlet. Using the metering and mixing techniques discussed above, these various reagents can be combined within reactors formed at the vertices of the matrix in order to accomplish reaction.

Embodiments of microfluidic matrix devices according to the present invention can also be utilized to screen compounds for a desired activity. With the devices described in FIGS. 28A-H for example, typical screening methods involve introducing a set of test compounds into the horizontal flow channels, with another set of compounds, cells, vesicles or the like being introduced via the vertical flow channels. Mixing occurs at the junctions and the presence or absence of the desired activity can then be monitored at vertices of the matrix.

For instance, a wide variety of binding assays can be conducted utilizing the microfluidic devices disclosed herein. Interactions between essentially any ligand and antiligand can be detected. Examples of ligand/antiligand binding interactions that can be investigated include, but are not limited to, enzyme/ligand interactions (e.g., substrates, cofactors, inhibitors); receptor/ligand; antigen/antibody; protein/protein (homophilic/heterophilic interactions); protein/nucleic acid; DNA/DNA; and DNA/RNA. Thus, the assays can be used to identify agonists and antagonists to receptors of interest, to identify ligands able to bind receptors and trigger an intracellular signal cascade, and to identify complementary nucleic acids, for example. Assays can be conducted in direct binding formats in which a ligand and putative antiligand are contacted with one another or in competitive binding formats well known to those of ordinary skill in the art. Binding assays can be conducted in heterogenous formats, as well as homogenous formats. In the homogeneous formats, ligands and antiligands are contacted with one another in solution and binding complexes detected without having to remove uncomplexed ligands and antiligands. Two approaches frequently utilized to conduct homogenous assays are fluorescence polarization (FP) and FRET assays.

Immunological assays are one general category of assays that can be performed with certain of the microfluidic devices disclosed herein. Some assays are conducted to screen a population of antibodies for those that can specifically bind to a particular antigen of interest. In such assays, a test antibody or population of antibodies is contacted with the antigen. Typically, the antigen is attached to a solid support. Examples of immunological assays include enzyme linked immunosorbent assays (ELISA) and competitive assays as are known in the art.

A variety of enzymatic assays can be performed using some of the devices disclosed herein. Such enzymatic assays generally involve introducing an assay mixture containing the necessary components to conduct an assay into a flow channel or junction for reaction with an enzyme that is subsequently introduced. The assay mixtures typically contain the substrate(s) for the enzyme, necessary cofactors (e.g., metal ions, NADH, NADPH), and buffer, for example. If a coupled assay is to be performed, the assay solution will also generally contain the enzyme, substrate(s) and cofactors necessary for the enzymatic couple.

A number of different cell reporter assays can be conducted with the provided microfluidic devices. One common type of reporter assay that can be conducted include those designed to identify agents that can bind to a cellular receptor and trigger the activation of an intracellular signal or signal cascade that activates transcription of a reporter construct. Such assays are useful for identifying compounds that can activate expression of a gene of interest. Two-hybrid assays, discussed below, are another major group of cell reporter assays that can be performed with the devices. The two-hybrid assays are useful for investigating binding interactions between proteins.

Cells utilized in screening compounds to identify those able to trigger gene expression typically express a receptor of interest and harbor a heterologous reporter construct. The receptor is one which activates transcription of a gene upon binding of a ligand to the receptor. The reporter construct is usually a vector that includes a transcriptional control element and a reporter gene operably linked thereto. The transcriptional control element is a genetic element that is responsive to an intracellular signal (e.g., a transcription factor) generated upon binding of a ligand to the receptor under investigation. The reporter gene encodes a detectable tran-

scriptional or translational product. Often the reporter (e.g., an enzyme) can generate an optical signal that can be detected by a detector associated with a microfluidic device.

In addition to the assays just described, a variety of methods to assay for cell membrane potential can be conducted with the microfluidic devices disclosed herein. In general, methods for monitoring membrane potential and ion channel activity can be measured using two alternate methods. One general approach is to use fluorescent ion shelters to measure bulk changes in ion concentrations inside cells. The second general approach is to use of FRET dyes sensitive to membrane potential.

Assays of cell proliferation can also be monitored with certain of the microfluidic devices disclosed herein. Such assays can be utilized in a variety of different studies. For example, the cell proliferation assays can be utilized in toxicological analyses, for example. Cell proliferation assays also have value in screening compounds for the treatment of various cell proliferation disorders including tumors.

The microfluidic devices disclosed herein can be utilized to perform a variety of different assays designed to identify toxic conditions, screen agents for potential toxicity, investigate cellular responses to toxic insults and assay for cell death. A variety of different parameters can be monitored to assess toxicity. Examples of such parameters include, but are not limited to, cell proliferation, monitoring activation of cellular pathways for toxicological responses by gene or protein expression analysis, DNA fragmentation; changes in the composition of cellular membranes, membrane permeability, activation of components of death-receptors or downstream signaling pathways (e.g., caspases), generic stress responses, NF-kappaB activation and responses to mitogens. Related assays are used to assay for apoptosis (a programmed process of cell death) and necrosis.

By contacting various microbial cells with different test compounds, some of the devices provided herein can be used to conduct antimicrobial assays, thereby identifying potential antibacterial compounds. The term "microbe" as used herein refers to any microscopic and/or unicellular fungus, any bacteria or any protozoan. Some antimicrobial assays involve retaining a cell in a cell cage and contacting it with at least one potential antimicrobial compound. The effect of the compound can be detected as any detectable change in the health and/or metabolism of the cell. Examples of such changes, include but are not limited to, alteration in growth, cell proliferation, cell differentiation, gene expression, cell division and the like.

Additional discussion of biological assays that can be conducted with certain of the microfluidic devices disclosed herein is provided in commonly owned PCT application PCT/US01/44869, filed Nov. 16, 2001.

Beyond uses in performing multiple independent assays, embodiments of microfluidic matrix devices in accordance with the present invention could also be used for multi-step assays. For example, after the reactions of the first-step are completed in the formats demonstrated above, the products may be isolated into the central compartments of each of the individual reactors. The other compartments may then be filled with the reagents for the downstream assays by operating the valve systems exactly in the same way as before. Therefore, the intermediate products need not be extracted off the chip, facilitating the automation of the multi-step assays and simplifying chip design and operation. Furthermore, this approach can save the unavoidable loss or dilution of product in the dead volume of the chip and the outlet port.

While the present invention has been described herein with reference to particular embodiments thereof, a latitude of

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modification, various changes and substitutions are intended in the foregoing disclosure, and it will be appreciated that in some instances some features of the invention will be employed without a corresponding use of other features without departing from the scope of the invention as set forth. Therefore, many modifications may be made to adapt a particular situation or material to the teachings of the invention without departing from the essential scope and spirit of the present invention. It is intended that the invention not be limited to the particular embodiment disclosed as the best mode contemplated for carrying out this invention, but that the invention will include all embodiments and equivalents falling within the scope of the claims.

What is claimed is:

1. A microfluidic device comprising:
 - a plurality of flow channels defined within an elastomer layer to form a matrix of rotary flow reactors;
 - a first set of control lines proximate to and separated from the flow channels by first elastomer membranes, the first elastomer membranes actuatable to introduce fluids into the rotary flow reactors and to isolate the rotary flow reactors; and
 - a second set of control lines proximate to and separated from the flow channels by second elastomer membranes, the second elastomer membranes actuatable to cause peristaltic pumping of the fluids within the isolated rotary flow reactors.
2. The microfluidic device of claim 1, wherein each rotary flow reactor comprises a plurality of column flow channels and a plurality of row flow channels from the plurality of flow channels.

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3. The microfluidic device of claim 2, wherein the plurality of column flow channels and the plurality of row flow channels of each rotary flow reactor comprises a first column flow channel, a second column flow channel, a first row flow channel, and a second row flow channel.

4. The microfluidic device of claim 3, wherein the first column flow channel and the second column flow channel of each respective rotary flow reactor are coupled with at least a third row flow channel forming another rotary flow reactor.

5. The microfluidic device of claim 4, further comprising a third set of control lines proximate to and separated from the flow channels by third elastomer membranes, the third elastomer membranes actuatable to isolate the first column flow channel from the second column flow channel of each respective rotary flow reactor.

6. The microfluidic device of claim 1, further comprising a common supply flow channel coupled with at least a subset of the plurality of column flow channels.

7. The microfluidic device of claim 1, further comprising a common waste flow channel coupled with at least a subset of the rotary flow reactors.

8. The microfluidic device of claim 1, further comprising a plurality of chamber structures, wherein each rotary flow reactor is coupled with at least one of the chamber structures.

9. The microfluidic device of claim 2, wherein at least a subset of the row flow channels from the plurality of flow channels each comprises a horizontal branch channel.

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